Chapter 10
The Interaction of Single Beams of X and Gamma Rays with a Scattering Medium

Radiation Dosimetry I

Text: H.E Johns and J.R. Cunningham, The physics of radiology, 4th ed.

http://www.utoledo.edu/med/depts/radther

1

Outline

- System of dosimetric calculations
  - TAR/TMR, Backscatter factor, PDD
- Effect of energy on photon beam dose deposition – PDDs and dose profiles
- Miscellaneous: equivalent square, blocking and scatter

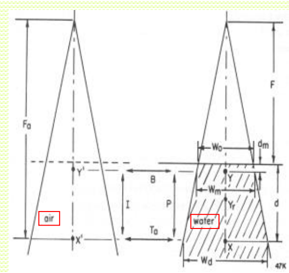
2

Dosimetric system

- Established a procedure for calculating dose at a point based on the measurement
- Now need to be able to calculate the dose at any point based on the known dose at the reference point
- A set of functions was developed to enable these calculations

3

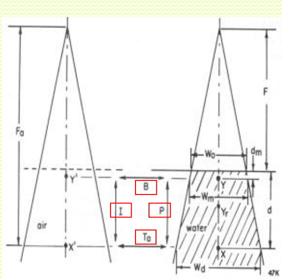
Parameters for calculation of absorbed dose



- Field width W
- Distance from the source F
- Depth in phantom d
- Depth of the maximum dose in phantom d_m
- Dose deposited at a certain point D_X, D_Y, etc.
- Dose is obtained under condition of electronic equilibrium (for air: enough phantom-like material surrounding the point)

4

Functions used in dose calculation



- Tissue-air ratio

  $T_a(d, W_d, hv) = D_X / D_{X'}$
- Backscatter factor

  $B(W_{m,} hv) =$

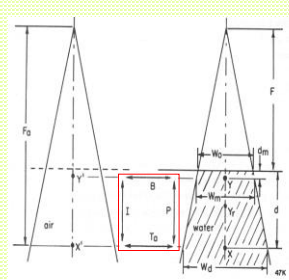
  $T_a(d_m, W_m, hv) = D_Y / D_{Y'}$
- Percentage depth dose

  $P(d, W_m, F, hv) = 100 D_X / D_Y$
- Inverse square law

  $I(F, d, d_m) = \dfrac{D_{X'}}{D_{Y'}} = \left(\dfrac{F + d_m}{F + d}\right)^2$

5

Relationship between functions

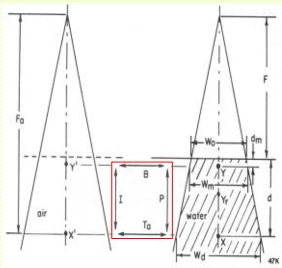


- Dose at one point can be calculated based on the known dose at a different point using a function along the corresponding arrow:

$$D_Y = D_{Y'} \cdot B$$
$$D_X = D_Y \cdot P$$
$$D_{X'} = D_{Y'} \cdot I$$
$$D_X = D_{X'} \cdot T_a$$

6

Relationship between functions



- More than one function can be involved in relating doses at different points:

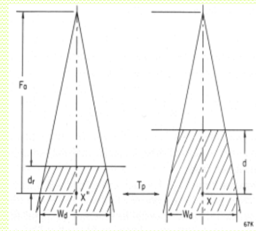
$$D_X = D_{Y'} \cdot B \cdot P$$
$$D_X = D_{Y'} \cdot I \cdot T_a$$

- Relationship between functions:

$$P(d, W_m, F) = \frac{100 \cdot T_a I}{B} = 100 \frac{T_a(d, W_d)}{B(W_m)} \left(\frac{F + d_m}{F + d} \right)^2$$

7

Tissue-phantom ratio



- For high-energy beams solid phantom is usually used for dosimetry
- Tissue-phantom ratio:

$$T_p(d, d_r, W_d, h\nu) = D_X / D_{X''}$$

- It can be related to T_a:

$$T_p = \frac{T_a(d, W_d, h\nu)}{T_a(d_r, W_d, h\nu)}$$

8

Tissue-air ratio

- Introduced to simplify calculations for rotational therapy with tumor located at the rotational axis
- In such arrangement the source-to-axis distance is fixed
- For distances larger than 50 cm T_a is independent of the distance to the source (first determined experimentally)

9

Tissue-air ratio

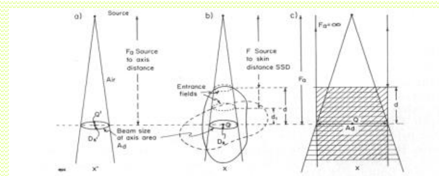


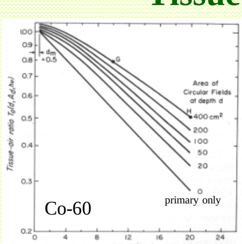
Figure 10-3. (a and b) Schematic diagram to illustrate the use of tissue-air ratio in dose calculations. (c) The scattering to point X from the cylindrical block of phantom material is the same as from the conical-shaped section when the two beams have the same area at depth d and receive the same primary radiation at X.

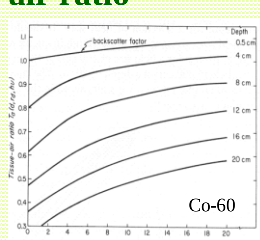
$$D_X = D_{X'} \cdot T_a(d, W_a, h\nu)$$

- T_a values are tabulated for different field sizes, depths, and energies (Table B-5d)

10

Tissue-air ratio





- Decreases almost exponentially with depth (Co-60 source is almost mono-energetic)
- Increases continuously with field size at all depth

11

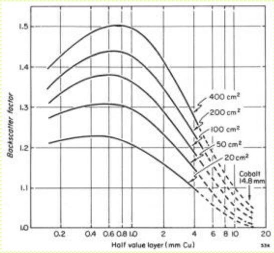
Backscatter factor

$$B(W_m, h\nu) = T_a(d_m, W_m, h\nu)$$

- Depends on the field size and quality of radiation
- Position of the maximum dose d_m depends on the field size and quality of radiation
- In general d_m is not the same as the depth where electronic equilibrium occurs: for large field sizes the scatter contribution determines the position of d_m

12

Backscatter factor



- The dependence on the radiation quality is non-monotonical, depends on field size
- For higher energies most of the scatter is forward-peaked: the amount of backscatter decreases

13

Percentage depth dose

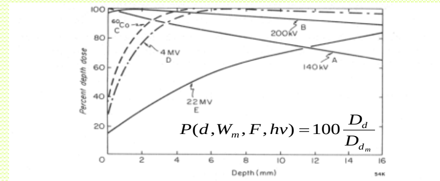


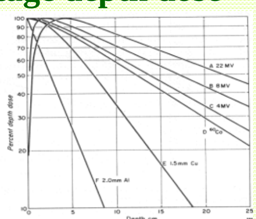
Figure 10-8. Percentage depth dose plotted against depth for the region near the surface for a range of photon beam energies.

- For higher energy beams there is a build-up region due to electrons scattered in forward direction
- After reaching its maximum dose deposition follows exponential attenuation

14

Percentage depth dose

Figure 10-9. Graph showing the variation of depth dose with depth: (a) 22 MeV radiation with copper compensating filter, 10 × 10 cm field, SSD 70 cm; (b) 8 MV radiation from a linear accelerator, 10 × 10 cm field, SSD 100 cm; (c) 4 MV radiation from a linear accelerator, 10 × 10 cm field, SSD 100 cm; (d) cobalt 60, 10 × 10 cm; (e) 200 kV, 10 × 10 cm field, HVL 1.5 mm Cu, SSD 50 cm; (f) 120 kV, HVL 2.0 mm Al. Area 100 cm², SSD 15 cm.



- After reaching its maximum dose deposition follows almost exponential decrease
- A useful quantity to use as an index of the penetration is the depth at which the dose falls to 50% of its peak value

15

Percentage depth dose

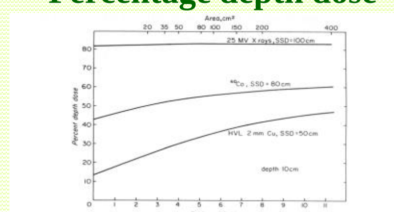


Figure 10-10. Variation of percentage depth dose with area and radius of field for three types of radiation. Depth 10 cm.

- For smaller filed sizes scatter contribution is small
- For high energy beams scatter contribution is small, therefore PDD is less dependent on the field size

16

Example 1

- All of the following are true regarding percentage depth dose (PDD) except:

A. Increases with increasing energy

B. Depends on field size

C. Is the dose at depth expressed as a percentage of the dose at d_m

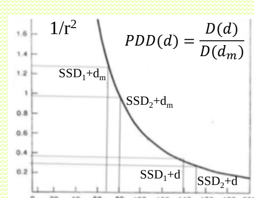
D. Decreases with increasing SSD

E. Decreases as depth increases

17

PDD with extended SSD

PDD increases with increasing SSD: ISL input decreases as distance increases through increase in SSD



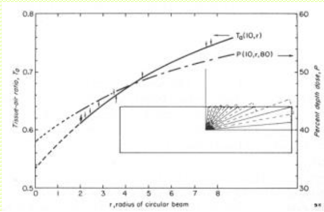
Mayneord's F Factor

$$F = \frac{(SSD_2 + d_{max})^2}{(SSD_1 + d_{max})^2}\frac{(SSD_1 + d)^2}{(SSD_2 + d)^2}$$

Quantitative example: for 6MV beam (d_m=1.5cm) changing SSD from 100 to 150cm will result in PDD(10cm) increase by ~5%

18

Equivalent field size



$$T_a(d, A) = \frac{1}{n}\sum_{i=1}^{n} T_a(d, r_i)$$

A – area of the field at depth d
r – radius of the circular beam

- Rectangular field can be divided into segments of circular fields
- Functions (T_a or P) can be calculated for irregular field sizes
- Smaller angular segments increase the accuracy (10^o is typically enough)

19

Equivalent field size

- A rectangular field gives a smaller depth dose and tissue-air ratio than does a circular or square field of the same area
- Rule of thumb method: Square fields and rectangular fields are equivalent if the ratios formed by dividing area by perimeter are the same (a/p)
- Tables 10.3 and 10.4 show radii of circular field and side length of square field corresponding to equivalent rectangular fields

20

Equivalent field size

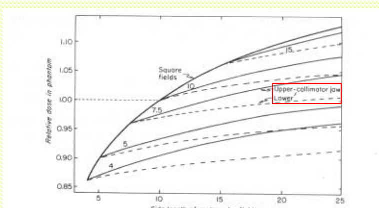


Figure 10-13. Relative output data for a 25 MV linear accelerator. The outputs were measured at a depth of 7 cm in a water phantom and are expressed as fractions of the output for a 10 × 10 cm field.

- In practice the output of the machine needs to be carefully measured for rectangular field sizes, due to the scatter from accelerator head

21

Example 2

- A field measuring 5 x 25 cm at SSD has an "equivalent square" field of side ____cm.

A. 5.0
B. 8.3
C. 11.2
D. 16.1
E. 25.0

$$\frac{a \cdot b}{2(a+b)} = \frac{c^2}{4c}$$

$$c = 2\frac{a \cdot b}{a+b} = 2\frac{5 \cdot 25}{5+25} = 8.3$$

22

Dose profile

Geometric penumbra at depth *d*

$$p = \frac{s*(SSD+d-f_c)}{f_c}$$

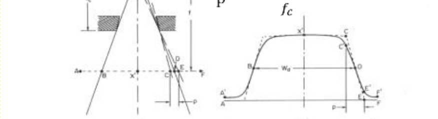


Figure 10-22. Diagrams illustrating properties of the primary component of a radiation beam. (a) The geometrical factors that lead to beam penumbra. (b) A dose profile as measured in air along the line A-F of a.

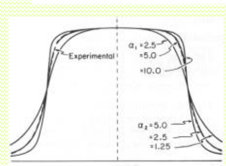
- Finite source size and beam shaping devices introduce geometric penumbra for primary radiation profile
- Scatter and flattening filter soften the edges of the beam

23

Dose profile

- For field size (beam width) at depth d, W_d, geometrical penumbra width *p*, the beam profile can be describe by a function:

$$f(x) = 1. - 0.5\, e^{(-\alpha_1/p)[(W_d/2) - |x|]} \text{ for } |x| \le w_d/2$$
$$f(x) = t + (0.5 - t)\, e^{(-\alpha_2/p)[|x| - (W_d/2)]} \text{ for } |x| > w_d/2 \qquad (10\text{-}16)$$

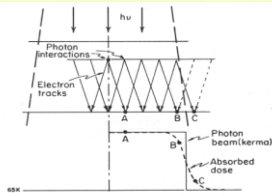


Parameters α_1, α_2, and t are determined by best fit of measured profiles

24

Dose profile

Figure 10-25. Diagram depicting loss of electronic equilibrium near the edge of a high energy photon beam. Point A receives electron tracks coming from all directions. Point B is inside the photon beam but receives fewer electron tracks than does A. Point C is outside of the photon beam but still receives some electron tracks.



- Even in the absence of scatter from beam shaping devices (primary collimator, jaws, etc.) the absorbed dose profile has a significant penumbra region
- Loss of lateral electronic equilibrium between kerma (energy lost by photons) and absorbed dose (energy lost by electrons)

25

Description of the radiation beam

- Primary and scatter components: zero-field tissue-air ratio and scatter-air ratio

$$T_a(d, r_d, h\nu) = T_a(d, 0, h\nu) + S(d, r_d, h\nu)$$

- For high-energy beams tissue-air ratio is replaced with tissue-phantom ratio

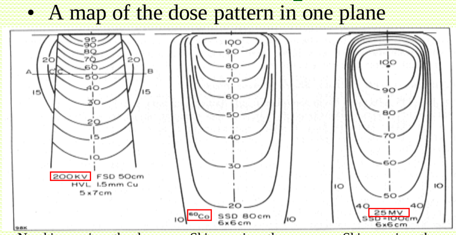
$$T_a(d, r_d, h\nu) = T_p(d, d_r, r_d, h\nu) \cdot T_a(d_r, r_d, h\nu)$$

- Due to relatively small amount of scattered radiation at high energies, $T_a(d_r, r_d, h\nu) \sim 1$

26

Isodose curves - photons

- A map of the dose pattern in one plane



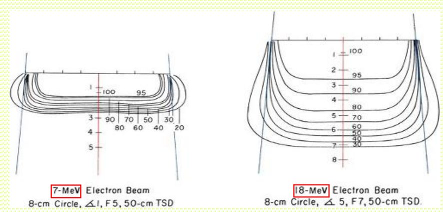
- No skin sparing, the dose falls continuously
- 25% at 10 cm depth
- Sharp beam edges (small source size)
- Large amount of side scatter (beyond the beam)

- Skin sparing, the maximum at 5 mm
- 52% at 10 cm depth
- Penumbra
- Low side scatter

- Skin sparing, the maximum at 4 cm
- 83% at 10 cm depth
- Penumbra
- Low side scatter

27

Isodose curves - electrons



- Almost no skin sparing, the maximum is close to the surface
- Bulging for low dose lines
- Constriction for high dose lines

28

Example 3

- Compared with 6 MeV electrons, superficial x-rays:

A. Have a lower skin dose
B. Deliver less dose to underlying tissues
C. Require thicker shielding
D. Have a sharper penumbra

29

Example 4

- In irregular field calculations, the increase in MU setting to account for blocking is greatest for:

A. 18 MV photons, 12 cm depth
B. 18 MV photons, d_m
C. 6 MV photons, 12 cm depth
D. 6 MV photons, d_m

Where the scatter contribution is the greatest: lower energy, greater depth

30

Radiation beam characterization

- TG-106 report on accelerator beam commissioning: at a minimum, the following data should be collected during commissioning:
  - For photon beams—percent depth dose PDD and profiles in-plane and/or cross-plane at various depths for open and wedge fields, data related to multileaf collimator
  - MLC such as inter- and intraleaf leakage, penumbra, tongue and grove effect, etc., head collimator scatter, total scatter, tray, and wedge factors
- TPS guidelines prescribe data to be collected

I. J. Das, T. C. Zhu, et al. "Accelerator beam data commissioning equipment and procedures: Report of the TG-106 of the Therapy Physics Committee of the AAPM," Med. Phys. **35**, 4186, 2008.

31

Radiation beam characterization

- TG-21 report – protocol on clinical reference dosimetry, based on air kerma (obsolete)
- TG-51 report with addendum – protocol on clinical reference dosimetry, based on dose to water
- TG-40 report - protocol on comprehensive QA for medical linear accelerators
- TG-142 report - the most updated protocol on comprehensive QA for Radiation therapy, includes QA for on-board imaging and S(B)RT

P.R. Almond et al., "AAPM's TG-51 protocol for clinical reference dosimetry of high-energy photon and electron beams", Med. Phys. **26**, 1847-1870, 1999; M. McEwen et al. Addendum to the AAPM's TG-51 protocol for clinical reference dosimetry of high-energy photon beams. Med Phys. 2014;41:041501.
G. J. Kutcher et al., "Comprehensive QA for radiation oncology: Report of AAPM radiation therapy committee task group 40," Med. Phys. **21**, 581–618, 1994.
E.E. Klein, et al. "Task Group 142 report: Quality assurance of medical accelerators," Med. Phys. **36**, 4197-4212, 2009.

32

Summary

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- Effect of energy on photon beam dose deposition – PDDs and dose profiles
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33