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A modified formula for defining tissue phantom ratio of photon beams

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Abstract

Tissue phantom ratio (TPR), for square fields of various dimensions has been determined at varying depths in water. The dose in water has been measured at a fixed source-to-surface distance (SSD) of 100 cm and reference depth of 5 cm for 6 MV photon beam of Siemens Linear Accelerator Primus II in German Cancer Research Center (DKFZ), Heidelberg, Germany. A modified formula has been developed to calculate the TPR value for isocentric treatment. The present article describes the conversion of the measured data values into a comprehensive and consistent data set by the modified formula, that gives the TPR from Percentage Depth Dose (PDD) with depth as a function of field sizes from 10 mm x 10 mm upto 300 mm x 300 mm) and depth (from 0 mm to 300 mm).

Introduction

The Tissue Air Ratio (TAR) concept works well in iso-centric setup for photon energies of cobalt-60 and below. For mega-voltage x-rays, produced by high energy medical linear accelerator - the TAR concept breaks down, because of difficulties in measuring the "dose to small mass of water in air" at those energies (the size of the required buildup cap for the ionization chamber becomes excessively large). To bypass this problem, the concept of tissue phantom ratio (TPR) was introduced for use in mega-voltage iso-centric setup. For the measurement of depth dose in SSD and SAD method in water and in air ---the PTW water phantom with MP3 interface has been used to drive the Farmer ion chamber of volume 0.01cc. Mephysto software was also used to drive the ion chamber for data acquisition of increment of 1mm both laterally and in depth. Sigma Plot 8.0 was adopted to analyze the measured data, and calculate the standard deviation[1, 2].

Normally TPR value was calculated from the measured PDD data without any modification $TPR_{20,10}$ can also be obtained from the simple relation:

$TPR_{20,10}$=1.2661 $PDD_{20,10}$ - 0.0595 ……………(1)

Here $PDD_{20,10}$ is the ratio of the percent depth-doses at 20 cm and 10 cm for a field size of 10 cm x 10 cm defined at the phantom surface with an SSD of 100 cm. The tissue-phantom ratio is formed by the ratio of the absorbed dose D_2 at depth d to the dose D_1 at reference depth d_r (Fig. 1).

TPR(d,B)= D_2/D_1…………………………………(2)

It is important to note that in this formulation, the field size refers to the field size at depth d. Accordingly, D_3/D_4 = TPR (d,C). The tissue-phantom ratio depends on fewer parameters than does the percentage depth dose. In particular, in this definition TPR is independent of the distance from the source.

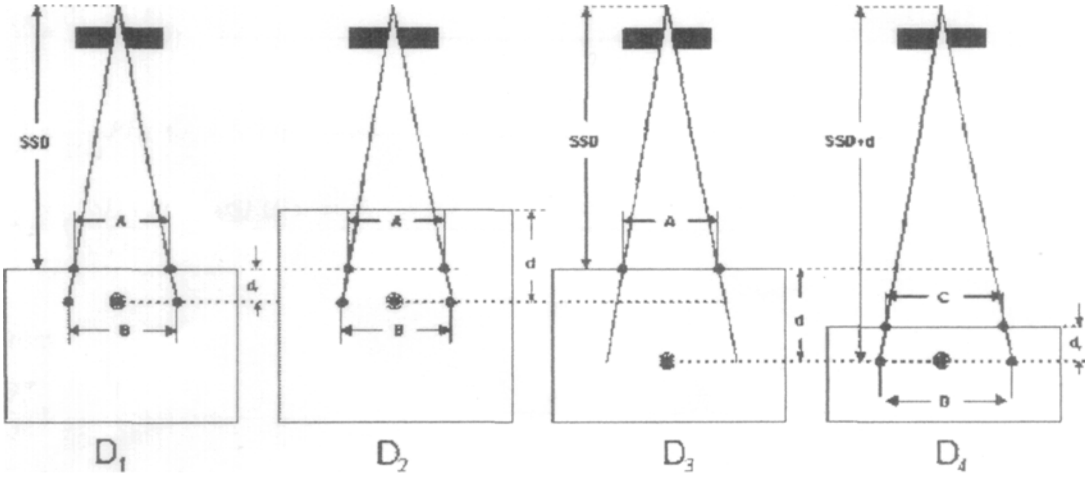

Figure 1: Schematic diagram of the measurement setup to obtain PDD and TPR

In identical system the TPR measurement is carried out using the following formula TPR (z, A_Q, h_γ) = Dose rate at P point (D_p)/Dose rate at P_{ref} = Dp/D_{pref}.

Here D_p is measured at an arbitrary depth, where we want to know the TPR value and D_{ref} is the measured dose at a reference depth of typically 5 cm on the central beam axis. It is important to mention here that the field size is always constant at point Q in both the diagrams. The diagram shows that we need to measure the dose at the two different conditions. We need to setup the dosimetry procedure twice for the measurement, thus it has a chance to make an error [2,3]. One is due to the setup and another is while filling up or removing water from the water phantom. This is also time consuming. In the work, a modified formula has been developed to measure the TPR value of any field size at any depth from PDD values [4,5]. The main advantage of this formulation is that, we can easily measure PDD of any field size at any depth. It has less chance to make mistake, because the ion chamber placement is done once and that is at the central axis of the beam.

The Formulation

The diagram [Fig.1] of the measurement schemes for PDD and TPR in a water phantom has illustrated their relationship.

$$\frac{D_2}{D_1} = TPR(d, B) \qquad (3)$$

$$\frac{D_3}{D_4} = TPR(d, D) \qquad (4)$$

$$\frac{D_3}{D_1} = \frac{1}{100} PDD(d, A, SSD) \qquad (5)$$

$$\frac{D_4}{D_1} = \left(\frac{SSD + dr}{SSD + d}\right)^2 \cdot \frac{PSF(C)}{PSF(A)} \qquad (6)$$

From

$$\frac{D_4}{D_1} = \frac{D_4}{D_3} = \frac{D_3}{D_1} \qquad (7)$$

*I*t follows

$$\left(\frac{SSD + dr}{SSD + d}\right)^2 \bullet \left(\frac{PSF(C)}{PSF(A)}\right) = \frac{\frac{1}{100} PDD(d, A, SSD)}{TPR(d, D)} \qquad ....$$

Using

$$B = A.\frac{SSD + dr}{SSD} \quad C = A.\frac{SSD + d - dr}{SSD} \quad D = A.\frac{SSD + d}{SSD}$$

One finally obtains

$$PDD(d, A, SSD) = 100.TPR(d.A\frac{SSD + d}{SSD}).\left(\frac{SSD + dr}{SSD + d}\right)^2 . \frac{PSF\left(A\frac{SSD + d - dr}{SSD}\right)}{PSF(A)} \qquad (10)$$

And

$$TPR(d, A\frac{SSD + d}{SSD}) = \frac{1}{100} PDD(d, A, SSD).\left(\frac{SSD + d}{SSD + dr}\right)^2 . \frac{PSF(A)}{PSF\left(A\frac{SSD + d - dr}{SSD}\right)} \qquad 11(a) \text{ Or}$$

$$TPR(d, A) = \frac{1}{100}.PDD\left(d, A\frac{SSD}{SSD + d}, SSD\right).\left(\frac{SSD + d}{SSD + dr}\right)^2 . \frac{PSF\left(A\frac{SSD}{SSD + d}\right)}{PSF\left(A\frac{SSD + d - dr}{SSD + d}\right)} \qquad 11(b)$$

A complete set of formulas is given to convert the PDD for photon beams from one SSD to another and to calculate the corresponding central axis quantities, and vice versa.

$$TPR(d, A) = \frac{1}{100}.PDD(d, A\frac{SSD}{SSD + d}, SSD.\left(\frac{SSD + d}{SSD + dr}\right)^2 . \frac{PSF(A\frac{SSD}{SSD + d})}{PSF(A\frac{SSD}{SSD + dr})} \qquad (12)$$

The reason for the difference may be due to the assumption used to derive equation 11, that the PSF is independent of SSD[6, 7, 8]. This is, however, in reality not true. For consistency reasons, the reader may be referred to the supplement 25 to the British Journal of Radiology if he wants to calculate TPR from PDD. Since the ratio of the peak scatter factors is often very close to unity, this ratio is frequently omitted. Then it does not matter whether equation 11 or 12 is used.

Results and Discussion

Figure 2 shows the calculated TPR (CTPR) values obtained by the modified formula versus depth in mm. Directly measured TPR values Vs depth in mm have been given in Figure 3. Information about the standard deviation of the measured and calculated TPR values are then given in Figure 4.

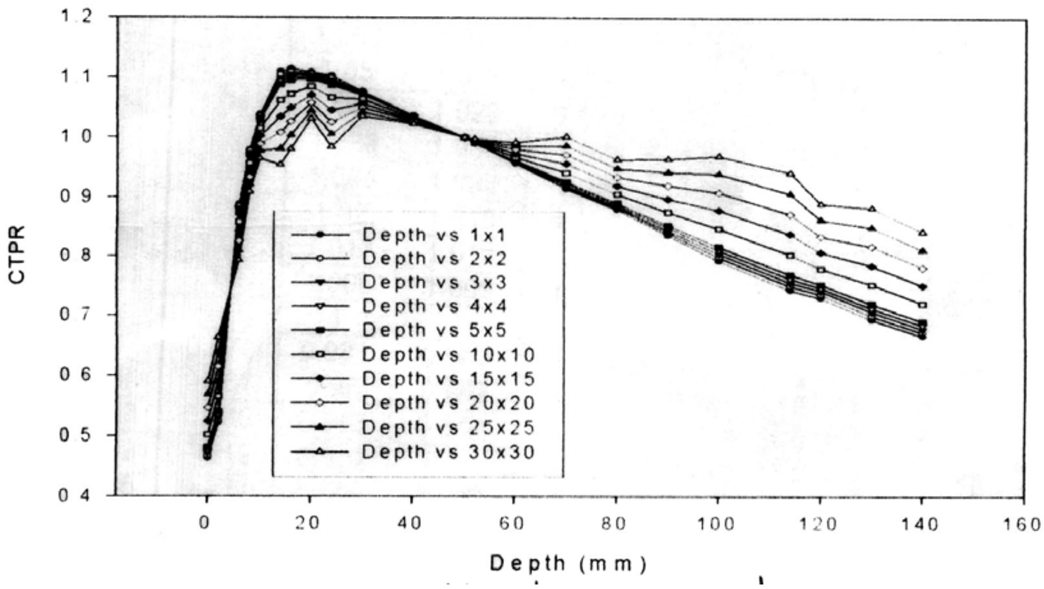

Figure 2: CTPR value obtained by modified formula Vs depth in mm

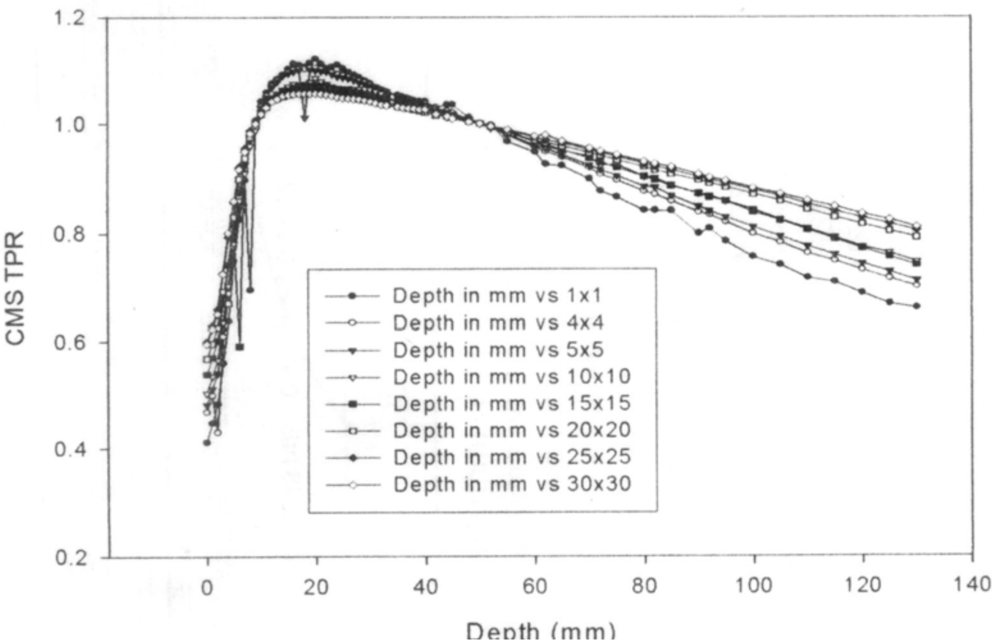

Figure 3: Directly measured TPR value Vs depth in mm

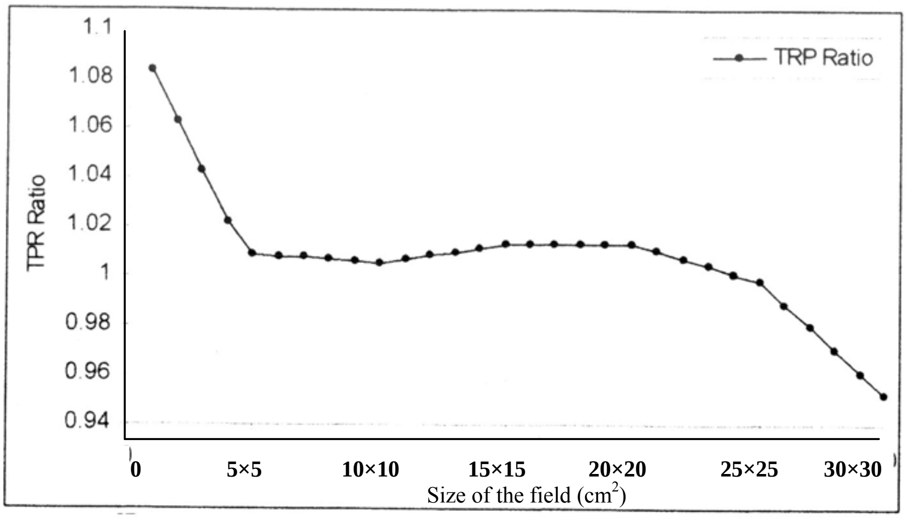

Figure 4: Ratio of standard deviation Vs corresponding field sizes

Several dosimetric functions are available to assist in the computation of absorbed dose in a patient based on the data measured in a phantom. The dosimetric functions describe the doses at various points in space "dressed" with a reference phantom, a phantom or a mini-phantom in air. Iso-centric measurement of ionizing radiation TAR is used only in the energy range of cobalt -60 gamma radiation and below, because of the difficulties of radiation measurement in air. So TPR is the value used in cobalt 60 gamma radiation and in any mega voltage photon beam radiation. It is possible to measure the TPR value in fixed SSD and in SAD formalism. In SAD formalism of measurement of TPR value, it is very difficult to measure it in various depths and for different field sizes. For direct measurement of TPR value of a linear accelerator we need to place the ionization chamber in air and in water at the different environments but in the same setup condition. Two points are worth mentioning here. One is regarding measurement of collimator scattering with mini phantom in air, which has a chance to make an error during setting up the dosimetry system for each and every field size[9, 10]. But it is possible to calculate the TPR value in direct measurement of fixed SSD formalism. In fixed SSD formalism PDD measurement is the easiest way and has less chance to make an error, because of the less difficulties in dosimetry setup. If we are able to measure the PDD value in reference condition then it is possible to calculate the TPR value of any field size in any depth.

For TPR data measurement we need to setup the ionization chamber in different conditions in air and in water. The setup will be varied for different field sizes with depth. So the whole procedure is time consuming and has the possibility to loss the consistency of measurement. But it is easy to calculate the TPR value of any field size of different depths in water. If we like to measure the PDD value of the reference condition or of any field size in depth, the formula can be adapted. Another point is that if we normalize the PDD data then it is not mandatory to correct the temperature and pressure.

In conclusion, the modified formulas are applicable to treat cancer, especially for treatment planning in isocentric procedure. This formula has been modified theoretically and is also verified experimentally. That’s why one needs to measure the data and put it by export file into the computerized treatment planning system for making a comparison of it by the newly developed formula. Some variation in the data is observed, but it is within the limit. This deviation of the measured and calculated data demands more research work to reduce the difference.

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