Analysis of Percentage Depth Dose for 6 and 15 MV Photon Energies of Medical Linear Accelerator With CC13 Ionization Chamber

M. A. Mia[1,2], M. S. Rahman[3*], S. Purohit[1], S. M. E. Kabir[4] and A. K. M. M. H. Meaze[1]

[1]*Department of Physics, University of Chittagong, Chittagong-4331, Bangladesh*
[2]*Chittagong Medical College and Hospital, Chittagong, Bangladesh*
[3]*Secondary Standard Dosimetry Laboratory (SSDL), Institute of Nuclear Science and Technology Bangladesh Atomic Energy Commission, Savar, Dhaka, Bangladesh.*
[4]*North East Medical College and Hospital, South Surma, Sylhet, Bangladesh*

Abstract

The success of external beam radiotherapy is mainly depended on how to commission and safety precaution is maintained before beam is heated to infected cells. CC13, an intermediate size ionization chamber is frequently used for the commissioning process of dose calculation for both standard and non-standard sizes field. The measurements of PDDs (Percentage of Depth Dose) were performed using TG-51 at the North East Cancer Hospital, Bangladesh using Varian Clinac IX-5982 with 6 MV & 15 MV photon beam energies for a set of 9 field sizes (4 × 4, 6 × 6, 8 × 8, 10 × 10, 15 × 15, 20 × 20, 25 × 25, 30 × 30 and 40 × 40 cm^2), keeping the same conditions such as pressure, temperature, incremental step, direction, geometry, chamber voltage, and polarity. The PDD for 6 & 15 MV photon beams were obtained with the above mentioned field sizes and at SSD 100 cm. The obtained PDD results showed that maximum dose depth (d_{max}) for above all mentioned field sizes were varied within 11.8 to 15.8 mm and 21.9 to 29.9 mm respectively for 6 and 15 MV photon energies. In modern clinical radiotherapy PDD is the most essential parameters for the commissioning of medical linear accelerator. Our measured values for depth dose (d_{max}) and PDD applying TG-51 protocol at 10 cm depth (d_{10}) for both 6 MV and 15 MV photon energies are found within the international refereed limits.

Keywords: Field size, percentage of depth dose, task group-51

1. Introduction

Since the inception of radiotherapy soon after the discovery of X-rays by Roentgen in 1895, the technology of X-ray production has first been aimed toward the treatment of injurious cells [1]. In modern radiation there are different types of techniques such as 3D Conformal Radiation Therapy (3D CRT), Intensity Modulated Radiation Therapy (IMRT) and Volumetric Modulated Arc Therapy (VMAT) allow for a more optimal dose of radiation against the target tumor and low doses of healthy tissue [2]. Therefore, for external radiation safety in radiation therapy determination of dosimetric characteristics of all radiation beams is vital so that the most appropriate set of treatment planning parameters is chosen. Data on the percentage depth-dose of diagnostic X-rays are important in evaluating patient dose from medical exposure. The quality of a radiation beam is usually expressed in terms of its penetrating power, which is a function mainly of the mean photon energy, and may be fully described by its depth dose characteristics in water but an increase in surface dose with field size is also noted due to electron scattering from intervening materials [3]. Data on dose distribution are almost entirely derived from measurements in phantoms, and then are used in a dose calculation system devised to predict dose distribution in an actual patient [4]. The production of radiation using sophisticated devices like LINAC is useful tools for clinical application. A linear accelerator customizes high energy x-rays to conform to a tumor's shape and destroy cancer cells while sparing surrounding normal tissue [5-6].

The LINAC can be used for therapy (treatment) after completion of some satisfactory scientific methods called as pre-commissioning testing and which is performed by a physicist by the process of optimization of treatment plan, and calculation of dose for certain plan to measure the dosimetric data. These parameters are important, firstly to use in radiotherapy treatment and secondly to evaluate and investigate physics of radiation beams [4, 6].

Medical linear accelerators (linacs) are cyclic accelerators which accelerate electrons to kinetic energies from 4 MeV to 25 MeV using non-conservative microwave RF fields in the frequency range from 10^3 MHz (L band) to 10^4 MHz (X band), with the vast majority running at 2856 MHz (S band). Various types of linacs are available for clinical use. Some provide x-rays only in the low megavoltage range (4 MV or 6 MV) others provide both x-rays and electrons at various megavoltage energies. A typical modern high energy linac will provide two photon energies (6 MV and 18 MV) and several electron energies (e.g., 6, 9, 12, 16, 22 MeV) [1].

Central axis dose distributions inside the patient or phantom are usually normalized to D_{max} = 100% at the depth of dose maximum z_{max} and then referred to as the percentage depth dose distributions.

The percentage depth dose is thus defined as follows:

$$PDD = 100\frac{D_Q}{D_P} = 100\frac{\dot{D}_Q}{\dot{D}_P} \quad (1)$$

Where, D_Q and $\dot{D}_Q$ are the dose and dose rate at point Q at depth z on the central axis of the phantom; D_P and $\dot{D}_P$ are the dose and dose rate at point P at z_{max} on the central axis of the phantom D [1]

*Corresponding author: *shakilurssdl@baec.gov.bd*

For megavoltage photon beams the dose which is generally much lower than the maximum dose and occurs at a depth z_{max} beneath the patient surface is known as surface dose [1]. The PDD and surface dose curve is shown in Fig. 1

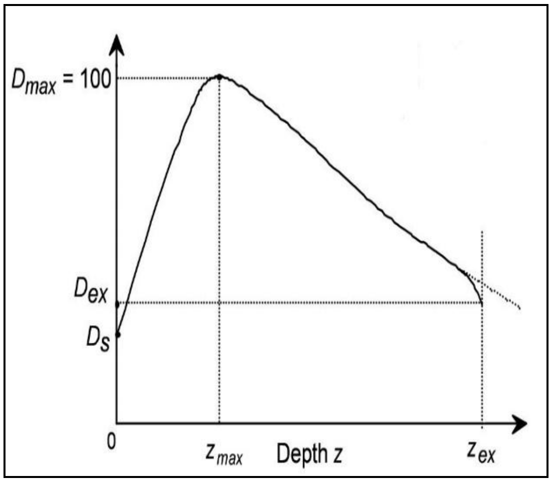

Fig. 1: For megavoltage photon beam, D_S is the surface dose at the beam entrance side, D_{ex} is the surface dose at the beam exit side. D_{max} is the dose maximum often normalized to 100 [1].

In this work, the basic and most important dosimetric parameter PDD for 6 and 15 MV photon beam energies with field size (s) 4 × 4, 6 × 6, 8 × 8, 10 × 10, 15 × 15, 20 × 20, 25 × 25, 30 × 30 and 40 × 40 cm^2 was measured. These fields were covered as the beam apertures of dimension from 4 × 4 to 40 × 40 cm^2 are generally used for delivering external photon beam treatment to patients [7].

2. Materials and Methods

2.1 Overview of the experimental facility

The Varian Clinac IX-5982 (Varian Medical System, Palo Alto, CA, USA) medical linear accelerator, having both photon and electron beam facility with dual energy mode has been used in this work. It has two photon energies e.g. 6 and 15 MV and several electron energies (e.g. 6 MeV, 9 MeV, 12 MeV, 15 MeV and 18 MeV). These measurements have been performed at the Department of Radiation Oncology of North East Cancer Hospital, South Surma, Sylhet by using CC13 (IBA Dosimetry GmbH, Schwarzenbruck, Germany) ion chamber (SN 14272) as a field chamber and CC13 (IBA Dosimetry GmbH, Schwarzenbruck, Germany) (SN 14273) as a reference chamber with a three dimensional (3D) computer controlled IBA EcoPhan water phantom (IBA Dosimetry GmbH, Schwarzenbruck, Germany, SN 1407) for the Varian Clinac IX-5982 (Varian Medical Systems, Palo Alto, CA, USA) medical linear accelerator (Model D-2300CD). More details of this Clinac are published elsewhere [8].

2.2 Experimental procedure

In this study, to measure percentage depth dose it is necessary to setup phantom and ionization chambers at isocentre alignment of the Varian Clinac IX-5982 medical linear accelerator system. In this step, the phantom and chambers were placed in isocentric distance of the LINAC having 6 and 15 MV photon beam energies. Before setup the ion chamber, three dimensional (3D) computer controlled IBA EcoPhan water phantom (SN 1407) i.e. water tank was leveled with spirit level and the source to water surface distance was set at 100 cm using the front pointer device. According to IAEA dosimetry protocol technical reports series, TRS-398 [9] and AAPM's protocol TG-51 [10], the effective point of measurement (EPOM) of a field chamber is shifted downstream by half of the chamber's inner radius (0.6r). During the measurement of the PDD an ionization chamber is placed at zero depth when the chamber's EPOM is aligned with water surface. This means that after the reference point has been temporarily aligned with water level, the chamber was moved downstream by the same distance, and zero set again. In this procedure, a reference chamber CC13 (SN 14273) was set in the corner of the measuring field just below the gantry where Fig. 2 shows the phantom setup for PDD measurements in Laboratory.

In the present work, to analyze the data for different FSs in 6 and 15 MV photon energies and examined the PDD characteristics, there are several important points to locate on the depth dose as (1) the surface dose or the dose at a depth of 0 cm; D_s, (2) the depth of dose maximum; $d_{max,}$(3) the dose at a depth of 10 cm; d_{10}, (4) depth of 50 % dose; D_{50}(%) and (5) average decrease in percent dose.

To measure the characteristics of the PDD for different fields size(s) the scanning was performed along central axis having measurement interval 2 mm starting from 300 mm depth to 0 mm depth in upward direction where the water turbulence throughout the phantom was avoided. The PDD curves were acquired for 9 square field sizes: 4 × 4, 6 × 6, 8 × 8, 10 × 10, 15 × 15, 20 × 20, 25 × 25, 30 × 30 and 40 × 40 cm^2 and where the above all field sizes were

Fig. 2: Phantom setup for PDDs measurements

defined by jaws, not multi-leaf collimator (MLC). As the room temperature evaporation can take place, therefore, during the PDD measurement the water level of 3D water phantom was always checked with the front pointer before the first scan and the water tank must be filled up with water in every 30 minutes to maintain the source to water surface distance constant. For this present study, the recorded data imported using Omni Pro-Accept 7.4c software (IBA Dosimetry, Germany) and the PDD curves were illustrated for 6 and 15 MV photon energies for LINAC system and finally the calculation and plotting the graph of this work was performed using MS Excel and Origin Pro software.

2.3 Measurement of percentage depth dose

To explore depth dose characteristics in diverse range of photon beam one way of characterizing the central axis dose distribution is to normalize dose at depth with respect to dose at a reference depth i.e. the central axis dose distributions inside the phantom or patient are usually normalized to D_{max}=100% at the depth of dose maximum $d_{max}$and then referred to as the percentage depth dose distributions.

Therefore, the PDD was determined by using the following equation [1]

$$PDD = \frac{D_Q}{D_P} \times 100\% \qquad (2)$$

Where, PDD is the percentage depth dose, D_Q is the absorbed dose at any point Q at the depth Z on the central axis of the phantom and D_P is the absorbed dose at point P at Z_{max}on the central axis of the phantom shown in Fig. 3.

3. Results and Discussion

The absorbed dose was described as PDD, which is a function of depth d, field size A and Source to Surface Distance (SSD) f. The dose at 0 cm depth i.e. the surface dose D_s (%) and the depth of 50% dose D_{50} (%) was displayed in Table 1.

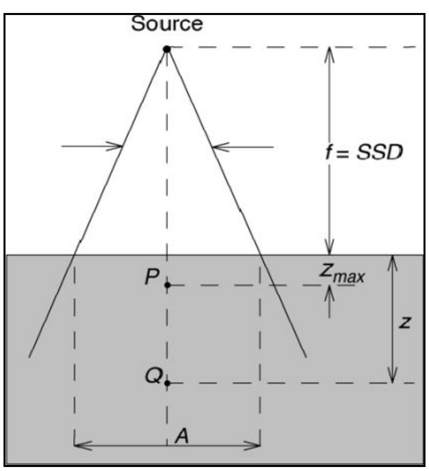

Fig. 3: Geometrical representation of the measurement of percentage depth dose (PDD) [9]

We experimentally found that percentage of depth dose increases with the increase of field sizes. Here, our measured doses showing better result than previous published work shown in Table 2. The average fall of doses (D_{max} to D_{50}) per cm decreases with the increase of field size(s) and beam energies given in Table 3.

Table 1: Determination of surface dose D_s (%) and depth of 50% dose D_{50} (%) for various field size(s)

Field Size (cm ^{2})	Values for 6 MV photon energy		Values for 15 MV photon energy	
	D_s(%)	D_{50}(%)	D_s(%)	D_{50}(%)
4 × 4	38.21	13.6	16.5	18.6
6 × 6	41.00	14.2	18.6	19.0
8 × 8	43.20	14.8	21.3	19.4
10 × 10	40.10	15.2	25.3	20.0
15 × 15	44.80	16.2	32.6	20.4
20 × 20	49.50	16.8	37.4	20.8
25 × 25	53.80	17.2	42.9	21.2
30 × 30	58.30	17.6	46.3	21.2
40 × 40	63.30	18.2	52.3	21.8

Table 2: Comparison of our measured surface dose (%) with previous published works

6 MV				15 MV		
Field size(s) (cm^2)	Measured surface dose (%)	Buzdar *etal.* [4]	Apipunyasopon *etal.* [11]	Fieldsize (s)	Measured surface dose (%)	Buzdar *etal.* [4]
10 × 10	40.1	52.4	55.0	10 × 10	25.3	28.5
15 × 15	44.8	55.1	60.0	15 × 15	32.6	33.9
20 × 20	49.5	58.6	65.1	20 × 20	37.4	38.8
25 × 25	53.8	61.3	-	25 × 25	42.9	43.4
30 × 30	58.3	64.4	-	30 × 30	46.3	47.6

Table 3: Average decrease of dose in the depth of d_{max} to d_{50}

Field size (cm^2)	Average decrease values	
	At 6 MV photon energy (% cm^{-1})	At 15 MV photon energy (% cm^{-1})
4 × 4	4.17	3.29
6 × 6	4.02	3.18
8 × 8	3.77	3.09
10 × 10	3.72	2.96
15 × 15	3.48	2.82
20 × 20	3.28	2.76
25 × 25	3.19	2.69
30 × 30	3.11	2.70
40 × 40	2.98	2.59

The average decrease of dose in standard field size (10 × 10) cm^2 has been measured 3.72% cm^{-1} and 2.96% cm^{-1} respectively for 6 and 15 MV photon energies. Our measured values were compared with previous published work in the standard field size (10 × 10 cm^2) for both 6 and 15 MV photon energies had the average decrease of doses (D_{max} to D_{50}) were 3.18% and 2.92% per cm respectively [4].

The measured depth of maximum dose (d_{max}) with 6 and 15 MV photon energies at 100 cm SSD in different field sizes and depth dose at 10 cm, d_{10} for standard reference field size (10 × 10 cm^2) with same photon energies are given in Table 4 and Table 5 respectively.

Table 4: Depth of dose maximum dose (d_{max}) of 6 and 15 MV photon energies for various field size(s)

Field size (cm^2)	d_{max}values	
	At 6 MV photon energy (cm)	At 15 MV photon energy (cm)
4 × 4	1.38	2.99
6 × 6	1.58	2.99
8 × 8	1.38	2.99
10 × 10	1.58	2.79
15 × 15	1.58	2.39
20 × 20	1.38	2.39
25 × 25	1.38	2.19
30 × 30	1.38	2.39
40 × 40	1.18	2.19

Table 5: Depth of dose maximum d_{max} and depth dose at 10 cm depth, d_{10} in 10 × 10 cm^2 field size for 6 and 15 MV photon energies

Beam energy (MV)	Field size (cm^2)	Depth dose d_{max} (mm)	PDD at 10 cm depth d_{10}(%)
6	10 × 10	15.8	66.8
15	10 × 10	27.9	76.8

The obtained PDD's of various field sizes showed that the maximum dose varies within 11.8 and 15.8 mm depth for 6 MV photon beam energy and whereas the maximum dose varies within 21.9 and 29.9 mm depth in 15 MV photon beam energy. For standard field size 10 × 10 cm^2 with 6 MV photon energy Yuliani*et al.* [2] obtained the maximum depth dose 14 mm.

The measured PDD (SSD = 100 cm) curves in all above mentioned FS(s) for 6 MV and 15 MV beam energies are respectively shown in Fig. 4(a) and 4(b). The PDD curves for reference field size and for 6 MV and 15 MV beam energies shown in Fig. 4(c).

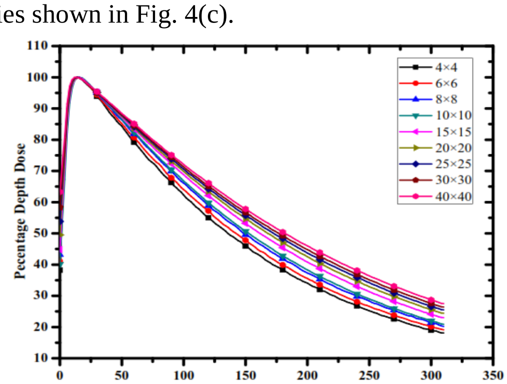

Fig. 4(a): PDD curve for all FS(s) with 6 MV photon beam energy

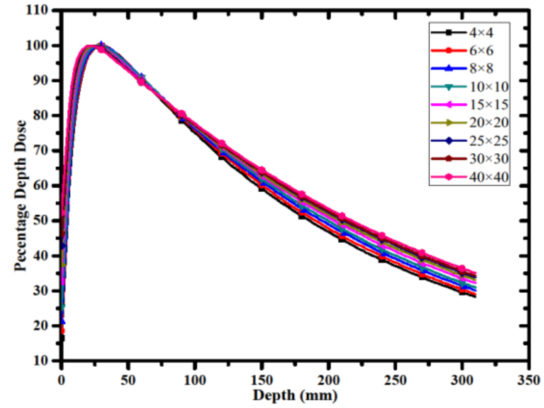

Fig. 4(b): PDD curve for all FS(s) with 15 MV photon beam energy

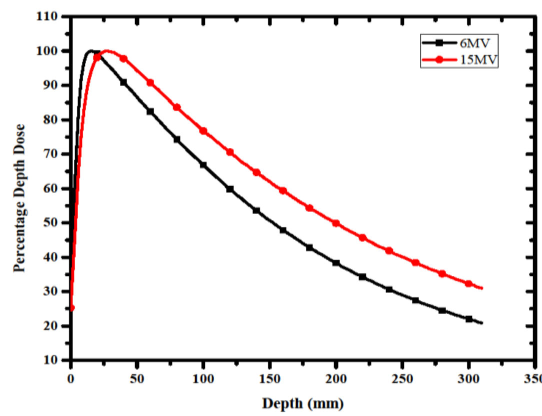

Fig. 4(c): PDD curve for (10 × 10) cm^2 FS with 6 MV and 15 MV photon beam energies

The depth of dose maximum (d_{max}) for field size 10 × 10 cm^2 for 6 and 15 MV photon energies are 15.8 mm and 27.9 mm respectively and the dose at 10 cm depth (d_{10}) were recorded as 66.8% and 76.8% respectively. The obtained doses for the depth of dose maximum (d_{max}) and at 10 cm depth (d_{10}) for both photon beam energies (6 and 15 MV) according to relative dosimetry are found within the limit mentioned by the American Association of Physicists in Medicines Task Group - AAPM TG51 [10].

According to AAPM TG-51, the tolerances for low-energy beams i.e. energy below 10 MV the depth of dose at 10 cm depth, $d_{max} \leq 75\%$ and for high energy $d_{max} \leq 89\%$ [10]and according to IEC 60731 scale [12], the tolerances for 6 MV and 10 MV photon energies for d_{max} are 1.5±0.2 cm and 2.3±0.2 cm, respectively.

In this study, for the quality assurance (QA) purpose, the American Association of Physicists in Medicine -Task Group-51 (AAPM TG 51) protocol [10] was followed for using CC13 ionization chamber. Although this CC13 IBA chamber was certified by PTB according to TRS 398 protocol but the measured depth of dose maximum d_{max} and dose at 10 cm depth, d_{10} for 6 MV and 15 MV photon beams were found within the limit mentioned in the several published works that were performed with TG-51. Depth of maximum dose is decrease with the increase of field sizes because surface doses are increase due to back scattering factors. Average fall of dose (d_{max} to d_{50}) per cm increase with the decrease of field sizes because in small field dose at penumbra is overlapped and sharp fall of dose in the penumbra region are not measured properly with intermediate size chamber [13]. From the Table 3, we can show that, the relationship between two energies is obvious, but the dose decrease rates do not have linear relationships although the FS (s) are same. The reason for non-linearity of dose decrease rates is due to the repeat mode of interaction with matter. As the higher energy beam interact with matter, with different attributes and hence its attenuation progression differ quite significantly from that of low energy beams.

4. Conclusion

For accurate radiotherapy treatment planning the percentage depth doses characteristics of medical LINAC are very important. By this study we can conclude that, dose decrease is a function of energy and higher energy beam have greater ability to penetrate. Due to other dosimetric considerations like FS and SSD, doses on certain locations can have different values, too, but the spectral and point to point distribution of the dose is the exclusive property of the beam energy. So, experimentally we have concluded that using intermediate size IBA chambers in different protocols rather than certified protocol in relative dosimetry all measured values lie in standard limit.

References

1. E. B. Podgorsak, Radiation Oncology Physics, A hand book for Teachers and Students, IAEA, Vienna, (2005).
2. W. S. Yuliani and H. S. Budi, Profile Dose and PDD Analysis in Small Photon Field with PTW Pinpoint Chamber 0.015 cc, Int. J. Inno. Res. Adv. Engr., **5**, 98-102 (2018).
3. M. Ravikumar and R. Ravichandran, Dose Measurements in the Build-up Region for the Photon Beams from Clinac-1800 Dual Energy Medical Linear Accelerator, Strahlenther Onkol., **176**, 223-8 (2000).
4. S. A. Buzdar, M. A. Rao and A. Nazir, An Analysis of Depth Dose Characteristics of Photon in Water, J. Ayub. Med. Coll. Abbottabad., **21**, 41-45 (2009).
5. D. J. Van and J. J. Battista, Cobalt-60: An Old Modality, A Renewed Challenge, **2** (1996).
6. R. N. Sruti, M. M. Islam, M. M. Rana, M. M. H. Bhuiyan, K. A. Khan, M. K. Newaz and M. S. Ahmed, Measurement of Percentage Depth Dose of a Linear Accelerator for 6 MV and 10 MV Photon Energies, Nucl. Sci. App., **24**, 29-32 (2015).
7. S. D. Sharma, Challenges of Small Photon Field Dosimetry are Still Challenging, J. Med. Phys., **39**, 131–132 (2014).
8. S. Purohit, S. M. E. Kabir, M. S. Rahman, M. K. A. Patwary, A. K. M. M. H. Meaze, I. Jahan, A. A. Mamun and D. Paul, A Study of Measurement of Relative Dose with Different Chambers for Small Radiotherapy 6MV Photon Field Dosimetry, Nucl. Sci. App., **26**, 23-27 (2017).
9. International Atomic Energy Agency, Technical Report Series (TRS) No. 398: Absorbed Dose Determination in External Beam Radiotherapy: An International Code of Practice for Dosimetry Based on Standards of Absorbed Dose to Water, Published by the IAEA on behalf of IAEA, WHO, PAHO, and ESTRO, IAEA, Vienna, 10A (2001).
10. P. R. Almond, R. Nath, P. J. Biggs, B. M. Coursey, D. W. Rogens, W. F. Hanson and M. S. Huq, AAPM’s TG-51 Protocol for Clinical Reference Dosimetry of High-Energy Photon and Electron Beams, Med. Phys., **26**, 1847-1870 (1999).
11. L. Apipunyasopon, S. Srisatitand N and Phaisangittisakul, An Investigation of the Depth Dose in the Build-up Region and Surface Dose for a 6-MV Therapeutic Photon Beam: Monte Carlo Simulation and Measurements, J. Rad. Res., **54**, 374-382 (2013).
12. International Electrotechnical Commission, Medical Electrical Equipment, Dosimeters with Ionization Chambers as Used in Radiotherapy, Standard IEC-60731 (1997).
13. A. J. D. Scott, A. E. Nahum and J. D. Fenwick, Using a Monte Carlo Model to Predict Dosimetric Properties of Small Radiotherapy Photon Fields, Am. Assoc. Phys. Med., **35**, 4671-4684 (2008).