Iranian Journal of Medical Physics
ijmp.mums.ac.ir

Suitability of Chambers in Relative Dose Measurement of Small Fields for Accelerated Photon Delivered by a Medical Linear Accelerator

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ARTICLE INFO

Article type:
Original Paper

Article history:
Received: Jan 30, 2020
Accepted: May 18, 2020

Keywords:
Clinical Protocols
Scattering Factors
Radiation Dosimetry
LINAC

ABSTRACT

Introduction: Using the small field in modern radiotherapy, the present study aimed at measuring the relative dosimetry (scattering factor, percentage depth dose (PDD), and profile of penumbra) with ionization (FC65-G, CC13, CC01) and diode (razor) chambers.

Material and Methods: Applying TRS-398 in Varian Clinac™ IX-5982 for 6 MV photon beams, the conditions (pressure, temperature, direction, polarity) were kept the same for a set of field sizes (1 × 1, 2 × 2, 3 × 3, 4 × 4, 5 × 5, 7 × 7, and 10 × 10 cm^2), and relative dosimetry was performed at the North-East Cancer Hospital, Sylhet, Bangladesh.

Results: During the output factor measurement in small fields, the razor showed better results than CC13. Taking CC01 as a standard in small fields, the data obtained from the study showed a good agreement with those of the previously published works.

Conclusion: Razor, with extremely small active volume, was very much suited for small field dosimetry, except for PDDs.

► *Please cite this article as:*
Purohit S, Meaze MH, Enamul Kabir SM, Rahman Sh, Patwary KA. Suitability of Chambers in Relative Dose Measurement of Small Fields for Accelerated Photon Delivered by a Medical Linear Accelerator. Iran J Med Phys 2021; 18: 293-299. 10.22038/ijmp.2020.46202.1721.

Introduction

Dosimetry of small fields in different radiotherapy poses is an increasingly challenging issue that exhibits a high degree of uncertainty, including both types A and B [1, 2, 3]. The new radiotherapy treatment planning, intensity-modulated radiotherapy (IMRT), image-guided radiotherapy (IGRT), and stereotactic radiosurgery (SRS) are included in external radiotherapy using a medical linear accelerator, which requires dosimetric characterization of small fields with correctly aligned detectors. Besides, the dosimetry of small irradiate fields is a challenging task that, due to the laws of the Bragg gray cavity theory, is sometimes not maintained by reference detectors [1]. Flounce perturbations are formed by some detectors with dimensions similar to those of small fields [1, 2]. Paskalev *et al.*, [1] made a conclusion on the percentage of depth dose measurement that a 1.5-mm diameter irradiation field made a 5% variation with 0.2 mm misalignment. It was concluded by Rikner [2] that placing orientation of an energy compensated diode chamber may cause serious practical problems in small field dosimetry. The usual detectors used to measure reference dosimetry (e g, RK cylindrical ion chamber, FC65-G) made them unfit when the field size was narrower than 2-3 cm. Actually, in such narrower fields, the lateral electronic equilibrium of the central axis does not properly hold, resulting in a peaked dose profile [1, 2]. Small geometric misalignment does not affect a larger field but, on average, affects the dose of the central axis by an amount, which actually varies according to the beam divergence effect depends on depth [3]. The well-known Monte Carlo modeling of linear accelerator established numerous codes that accurately model particle transport defining precise geometry and compositions of the various components. Heydarian *et al.,* [4] used fixed circular collimators of 5-23-mm diameter, applying EGS4 to model the output of a 6 MV Siemens Mevatron linear accelerator. They reported a good agreement between the Monte-Carlo-based percentage depth doses (PDDs) and values measured by a diamond detector. The measured PDD of their experiment with diode declined more sharply and with RK ion chamber PDDs lessening less stridently. Cheng *et al.,* [5] and

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Francescon *et al.,* [6] reported good agreement between Monte Carlo calculations and results obtained from diodes and micro-ionization chambers.

The current study aimed at investigating different active volume ionization and diode (razor) chambers to measure relative dosimetry and generalize a chamber better suited to small fields.

Materials and Methods

The relative absorbed dose to water was measured for small irradiation fields with the newly proposed formalism [7] using CC01, CC13, FC65-G (ionization), and razor (diode) chambers following the IAEA dosimetry protocol of TRS-398 [8] at reference conditions (100 cm source to surface distance (SSD), 10 cm depth in IAEA water phantom). The output factor (OF), cross line profile (CLP), the profile of penumbra (PP), and PDD for various small fields were measured following the methodology of published literature [2, 8]. The ionization (CC13, CC01, and FC-65G) chambers (iba dosimetry GmbH, Germany) were placed perpendicular to the field source, and the diode (razor) chamber was placed parallel to the source direction. In the present study, PDDs and profiles were measured with ionization chamber coupling with IBA electrometer Dose-1. The OmniPro-Accept version 7.4C (iba dosimetry GmbH, Germany) was connected to a radiation field analyzer via a computer.

The linear accelerator (Varian medical systems Deutschland GmbH, Germany) of model Varian Clinac™ *ix*-5982 [9] comprised of the facilities for volumetric modulated arc therapy (VMAT) and IMRT treatment techniques of North-East Cancer Hospital, Sylhet, Bangladesh, were employed to deliver 6 MV photon beams.

The three different ionization chambers, CC01, CC13, and FC65-G with 0.01, 0.13, and 0.65 cm^3 active volumes, respectively, were used for measurement. The central electrodes were made of steel in CC01, C552, and CC13 and aluminum in FC65-G chambers, and the walls of CC01, CC13, and FC65-G were made of C552, C552, and graphite, respectively.

The iba reference field diode RFD^{3G} and diode (razor) were the circles formed of geometric shapes of an active area with 2 and 0.6 mm diameters, respectively. The razor (diode) chamber had a nominal sensitivity of 4.1 nC/Gy calibrated in ^{60}Co beams. Energy dependence in 5 × 5 cm for 6 MV beams at 30 cm depth was maximum 1%. In 0.02 to 40 Gy conditions, dose linearity was typically <0.2% for absolute deviation from endpoint fit [10].

The echo water phantom-484841 (iba dosimetry GmbH, Germany) is a water phantom with three echo sensors on its X, Y, and Z axes and a remote pendant. A common control unit was integrated with the phantom and acted as an interface between the phantom and computer software, which allowed the ion and diode chambers to be remotely controlled. In the current study, the OmniPro-Accept version 7.4C software and a computer controller device were used for proper dosimetry. Several dosimetric parameters were calculated as follows:

Scattering factor

The total scattering factor in the dosimetry of the mega-voltage photon beam from the medical linear accelerator consists of two parts. First, the collimator scattering factor that contributes to a linear accelerator head and second, the phantom scattering factor. Holt *et al.*, [11] defined the total scattering factor (S_{cp}) in terms of the phantom (S_p) and collimator (S_c) scattering factors by Equation 1 given below, which is normalized to a 10 × 10-cm reference field:

$$S_{cp} = S_c \times S_p \qquad (1)$$

The phantom scattering factor can be defined as the ratio of the dose at the central axis in the clinical field to reference field:

$$S_p = \frac{D\,(f,S,D)}{D_{ref}\,(f,S_{ref},\ D)} \qquad (2)$$

S, *f*, and *d* represent the field size, source to surface distance, and depth, respectively. The phantom scattering factor in a fixed reference depth beyond d_{max} was recommended [12, 13], and a table of S_p values at 10 cm depth, between 4 × 4 and 40 × 40 cm fields, was established [14].

Output factor

It is known that the properties of human tissue are similar to those of water. Therefore, radiation prescribed for patients should typically be equivalent to doses absorbed by water during treatment planning system. OFs are generally measured by the ratio of monitor reading measured by different chambers at clinical and reference fields (10 × 10 cm) given in Equation 3.

$$Output\ factor = \frac{M_W\,(A,\ d)}{M_W\,(A_{ref},\ d)} \qquad (3)$$

where M_w represents monitor reading, *d* reference depth in a water phantom, and *A* and A_{ref} clinical and reference field sizes, respectively.

Central axis profile

The beam profile was measured for irradiated field size at a reference depth where the chamber was moved towards the cross line or in line throughout the field. Cross-beam profiles are sometimes not symmetric due to noncircular focal spot. Therefore, the cross beam profile is followed up with a set of two orthogonal dose profiles measured in phantom to the beam central line at a given depth. In the penumbra region, usually, 20%-80% dose variation could be observed.

Percentage of depth dose

PDD can be measured by Equation 4 as follows:

$$PDD = (D_d \div D_{dmax}) \times 100\% \qquad (4)$$

where D_d and D_{dmax} are the dose rate at any depth and maximum dose depth, respectively. In low-energy doses, usually up to 400 kVp, the depth of maximum dose is actually at the surface [15]. Beam quality, depth, field size and shape, source to surface distance, and beam collimation are the parameters affecting the central axis depth dose. As it is known, the range of production of secondary electrons increases with beam energy; therefore, the majority of small-field well-published papers, such as those of Verhaegen *et al.*, [15] and Heydarian *et al.*, [4] performed their work at 6 MV beams. Following their well-published works, the current study was also performed using 6 MV photon beams delivered by a Varian Clinac^{TM} *ix*-5982.

Results

The phantom scattering factor was measured using CC13 by applying Equation 2 at reference conditions in water phantom (depth 10 cm, SSD 100 cm) for various field sizes. The $TPR_{20,\ 10}$ value of the linear accelerator for 6 MV photon beams was 0.6670. The measured phantom scattering factors using the TRS-398 relative dosimetry protocol are given in Table 1**.**

The OF was measured using Equation 3, which is the ratio of doses by small non-standard to standard reference fields with various chambers. The measured OFs for four different chambers in varying field sizes are summarized in Table 2, and the comparison of the current study results with those of previous well-published works is shown in both Table 2 [16] and Table 3 [17].

Table1. Measurement of Phantom Scattering Factors in Different Small Fields

Field size (cm × cm)	S_p at 10-cm depth	S_p at D_{max}
1 × 1	0.6379	0.7275
2 × 2	0.7867	0.8740
3 × 3	0.8263	0.9120
4 × 4	0.8606	0.9288
5 × 5	0.8921	0.9468
7 × 7	0.9551	0.9739
10 × 10	1.0000	1.0000

Table 2. The Output Factors Normalized to Reference Field for Different Chambers with Field Sizes

Field size (cm^2)	The Study Measured OF[a]				Published OFs[16]	
	CC13	Razor	CC01	FC65-G	A16	EPID[b]
1 × 1	0.6378	0.6773	0.6616	0.3118	0.6860	0.7600
2 × 2	0.7865	0.7618	0.7755	0.6713	0.7820	0.7980
3 × 3	0.8262	0.8047	0.8156	0.8165	0.8240	0.8320
4 × 4	0.8604	0.8405	0.8522	0.8588	0.8580	0.8620
5 × 5	0.8919	0.8732	0.8859	0.8888	0.8900	0.8900
7 × 7	0.9549	0.9313	0.9352	0.9678	-	-
10 × 10	1.0000	1.0000	1.0000	1.0000	1.0000	1.0000

[a]OF: Output Factor, [b]EPID: Electronic Portal Imaging Devices

Table. 3 Relative Errors of Output Factors of CC13 and FC65-G Ionization Chambers When CC01 Taken as a Standard One

Field size (cm × cm)	Varian Linear Accelerator ix 5982 6 MV the Present Work		Varian 600C 6 MV Li Chen *et al.* [17]	
	CC13	FC65-G	CC13	FC65-G
1 × 1	3.60%	52.87%	14.47%	58.81%
2 × 2	1.42%	13.43%	0.58%	21.30%
3 × 3	0.13%	0.11%	0.29%	2.23%
4 × 4	0.96%	0.77%	0.53%	0.12%
5 × 5	0.68%	0.32%	0.48%	0.22%

Table 4. Using Different IBA Chamber Fall of Normalized Dose per Millimeter at Penumbra Region in Different Small Fields

Profile (cm^2)	CC01	CC13	Razor (Diode)
	Fall of Dose (%)/mm	Fall of Dose (%)/mm	Fall of Dose (%)/mm
1 × 1	18.2300	14.6400	24.8180
2 × 2	17.3005	10.7300	18.0863
3 × 3	15.8263	11.0823	17.3425
4 × 4	16.7078	10.9700	16.9238
10 × 10	9.8890	9.0643	13.9383

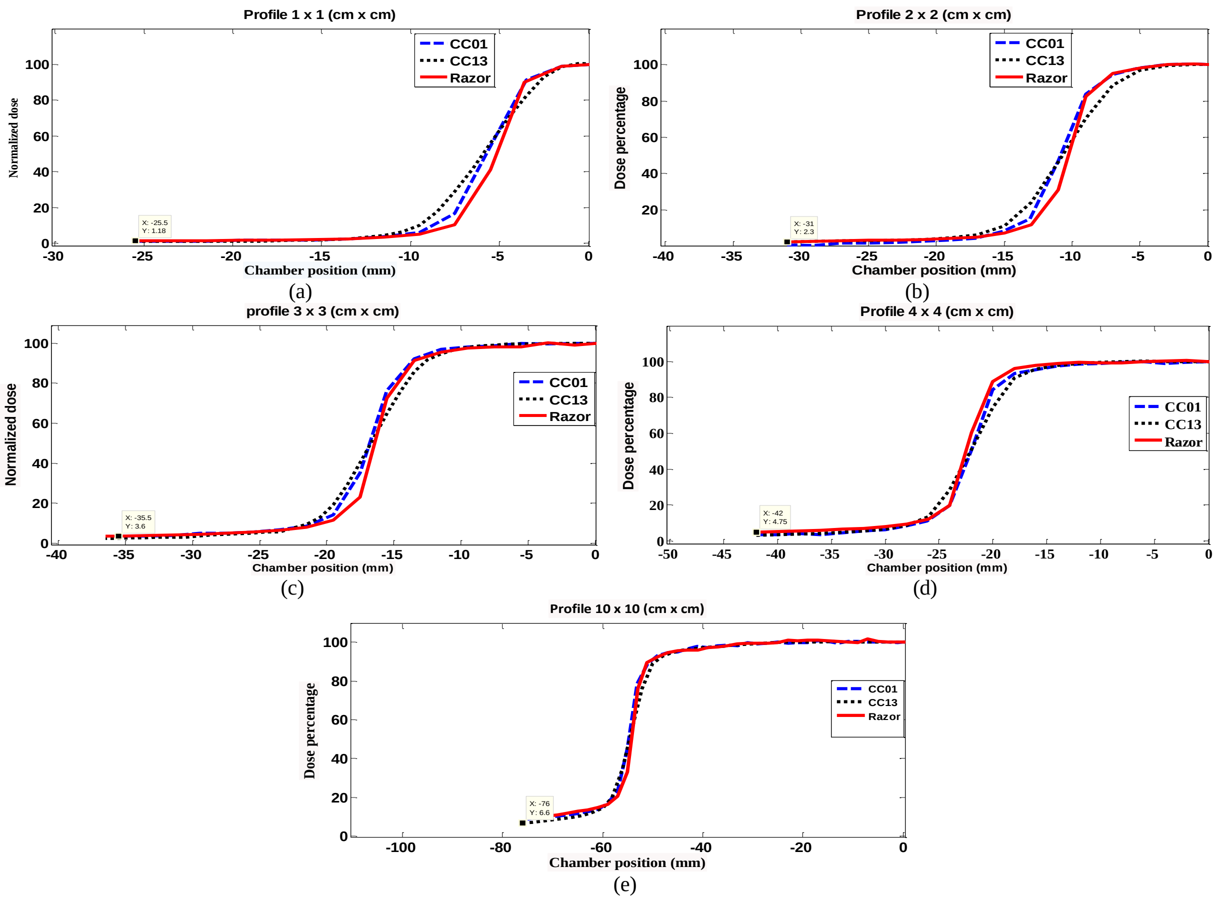
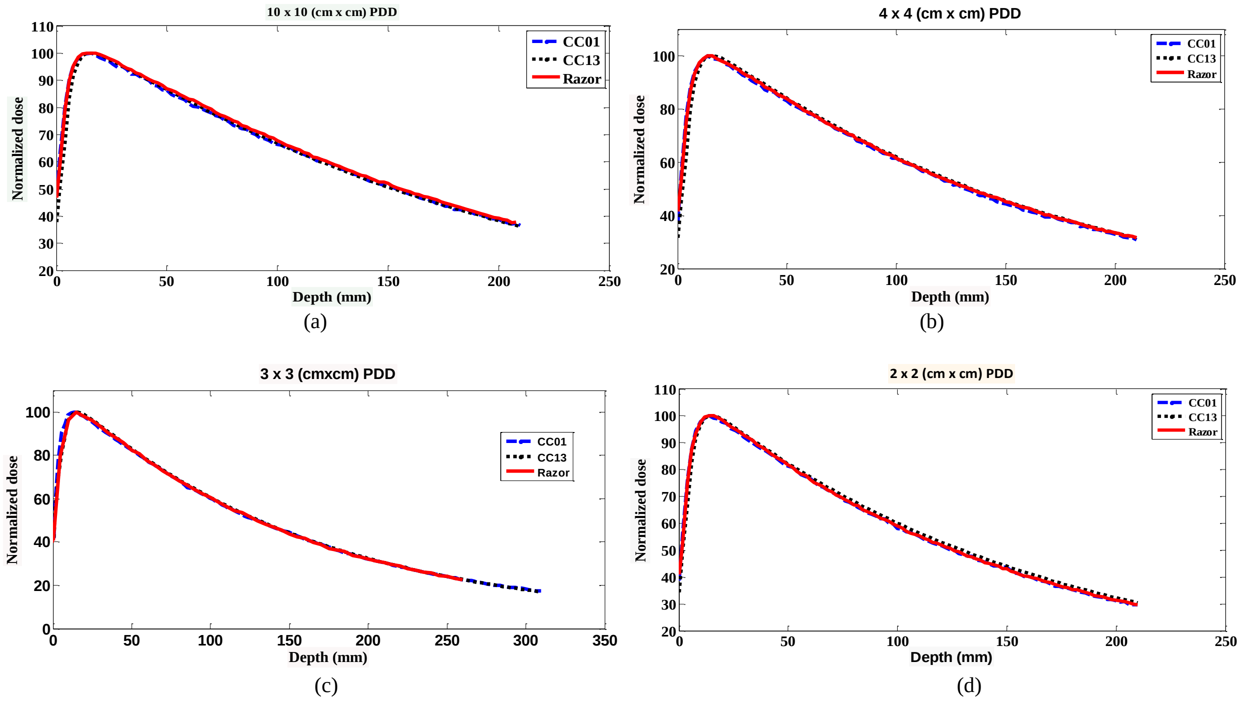
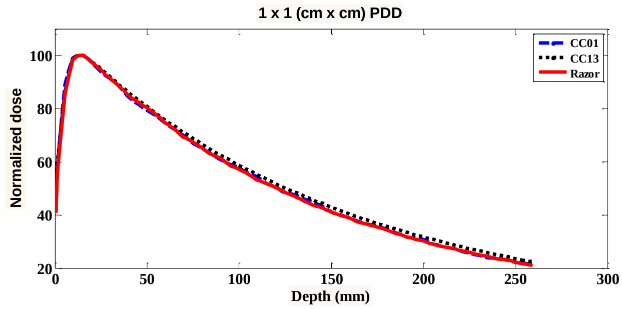


Figure 1. Representation of the cross line profile of different small and reference fields using three different chambers in relative dosimetry at 100 cm SSD and 10 cm depth





(e)

Figure 2. Experimental values of percentage of depth dose of reference and small fields at 100 cm SSD

Table 5. Measurement of Z_{max} from PDD Curves for Various Field Sizes Using Three Different Chambers

Field size (cm × cm)	Z_{max} in cm (CC01)	Z_{max} in cm (CC13)	Z_{max} in cm Razor(Diode)
1 × 1	1.35	1.35	1.35 1.55
2 × 2	1.38	1.39 1.58	1.39
3 × 3	1.55	1.55	1.48
4 × 4	1.38	1.59	1.38 1.58
10 × 10	1.58	1.58	1.58

Table 6. Percentage of Surface Dose Calculated Based on the Experimental Values of PDD

Field size (cm × cm)	Percentage of Surface Dose (CC01)	Percentage of Surface Dose (CC13)	Percentage of Surface Dose (Razor)
1 × 1	41.80	41.58	40.68
2 × 2	43.61	42.70	44.14
3 × 3	41.78	42.80	40.54
4 × 4	42.87	42.52	44.81
10 × 10	52.66	41.68	50.80

The CLP for small (1 × 1, 2 × 2, 3 × 3, 4 × 4 cm^2) and reference (10 cm × 10 cm) field sizes was measured using three different chambers by echo water phantom at 10 cm depth and 100 cm SSD. The beam profiles of penumbra and their variations with different ionization and diode chambers for different filed sizes are shown in Figure 1.

The fall of normalized dose (%) in penumbra region is calculated by:

= [*dose difference at penumbra* ($D_{80\%} - D_{20\%}$) ÷ *difference of chamber position* ($x_{80} - x_{20}$)].

In cross profile analysis, the experimental data in Figure 1 show that the dose of the central axis is scattered outward in all directions due to the overlapping of the penumbra. Since the dose of the central axis is going outward, the dose variation at the penumbra region is increased with the decrease of field size shown in Table 4. In the current experiment, the CC13 chamber showed that the broader penumbra is obtained in a smaller field than the CC01 and razor chamber.

The percentage of depth dose of 6 MV photon beam was measured for the field sizes of 10 × 10, 4 × 4, 3 × 3, 2 × 2, and 1 × 1 cm^2 at 100 cm source to surface distance, using a vertical IBA echo water phantom. According to different types of chambers, the percentage of depth doses for different field sizes is graphically represented in Figure 2. The maximum absorbed dose, Z_{max} in water is calculated based on the PDD curve tabulated and summarized in Table 5.

Table 5 shows that the value of Z_{max} depends on the field size. Table 6 also indicates that the percentage surface dose depends on field size.

Discussion

The razor (diode) chamber contains a non-water equivalent chip and a very small active volume. Hence, in the nearest jaws of collimator edges, the razor has more sensitivity to low-energy escalated doses. As a result, in the field sizes smaller than 3 × 3 cm, higher doses are measured in razor chambers compared to ionization ones. OF is the ratio of dose between small and reference fields (10 × 10 cm), so overestimation was observed in the experiment with a razor (diode) in a small field. The ionization chamber has a volume averaging effect and lower sensitivity to low-energy scattered doses that were extremely observed in the current experiment for the ionization chamber FC65-G.

In small field dosimetry, some active parts of a large volume chamber missed the irradiate field region. They resulted in a serious dose fall experimentally figured out at OF measurement. Alison Scott *et al.*, [3] measured OFs using a variety of detectors (pinpoint, diamond, shielded diode, unshielded diode, and film chamber), and their published data were in agreement with the current study measured OFs of ionization (CC01, CC13, FC65-G) and diode (razor) chambers. Tables 2 and 3 indicate differences between the results of the present and previous studies [16, 17], which could be due to different experimental methods, chambers, set-ups, and medical linear accelerators. During the OF measurement in the small field, the ionization chamber should be small enough with polarity correction to exclude the volume averaging effect [8, 18, 19]. The penumbra region of a profile has two components [20, 21], which has convolutions with photon fluence, dose distribution, and secondary electron distribution. The volume averaging effect of the intermediate volume ionization chamber obscures the penumbra of a small field in the profile measurement. Hence, isodose moves outward [22]. Therefore, for the ionization chamber CC13 in small field sizes, irradiate fields slightly increase. In the CC01 ionization chamber, Z_{max} decreases at 4 × 4 cm^2, which is due to the contribution of the scattering factor. For 2 × 2 cm^2, the range of Z_{max} is broadened for CC13 due to the participation of secondary charged particles. For the CC13 chamber, Z_{max} decreases from 4 × 4 cm^2 due to a very weak sensitivity to lateral scattering and uncertainty effect. In 1 × 1 and 4 × 4 cm^2 fields, Z_{max} was broadened using a razor (diode) chamber due to proper positioning and transient effect of a charged-particle equilibrium.

Conclusion

Razor (diode) has completely different water equivalence and a good response to lateral low-energy scattered doses. Therefore, in the normalization process with the reference field, quantum noise might also be normalized with dose. As a result, the overestimation of OF is visible for razor (diode). In measuring profiles and PDD curves, there is a lack of space of the plateau region for placing ionization chambers in the co-axial region of the small field. Therefore, it is difficult to place an ionization chamber inside a small field. Using razor (diode) chambers rather than ionization chambers smaller than 2 × 2 cm field sharply increased dose variation in the penumbra region**.** In the current study, the choice of razor (diode) chamber with a good response to the escalated dose in the penumbral region at a small field was a high-quality alternative. PDDs also increased with the enlargement of field sizes. According to the PDD analysis curve, using different active volume IBA chambers (CC01, CC13, and razor), percentage of surface dose increased with the increase of field size, except for CC13 with some discrepancies due to lack of charged-particle equilibrium. At 1 × 1 cm field, Z_{max} was broadened in the razor (diode) chamber. Therefore, it can be concluded that CC01 (ionization) with small active volume is a more suitable chamber for PDDs and OF than CC13 and razor.

Acknowledgment

This research work was performed in a collaboration with the University of Chittagong, North East Cancer Hospital Sylhet and Bangladesh Atomic Energy Commission Savar, Dhaka. We want to give thanks to Prof. Dr. Syed Akram Hussain the head of the department of Oncology in North East Cancer Hospital, Sylhet for providing Laboratory facilities.

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