Implementation and Challenges of International Atomic Energy Agency/American Association of Physicists in Medicine TRS 483 Formalism for Field Output Factors and Involved Uncertainties Determination in Small Fields for TomoTherapy

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Abstract

Purpose: International Atomic Energy Agency published TRS-483 to address the issues of small field dosimetry. Our study calculates the output factor in the small fields of TomoTherapy using different detectors and dosimetric conditions. Furthermore, it estimates the various components of uncertainty and presents challenges faced during implementation. **Materials and Methods:** Beam quality $TPR_{20,10}(10)$ at the hypothetical field size of 10 cm × 10 cm was calculated from $TPR_{20,10}(S)$. Two ionization chambers based on the minimum field width required to satisfy the lateral charge particle equilibrium and one unshielded electron field diode (EFD) were selected. Output factor measurements were performed in various dosimetric conditions. **Results:** Beam quality $TPR_{20,10}(10)$ has a mean value of 0.627 ± 0.001. The maximum variation of output factor between CC01 chamber and EFD diode at the smallest field size was 11.80%. In source to surface setup, the difference between water and virtual water was up to 9.68% and 8.13%, respectively, for the CC01 chamber and EFD diode. The total uncertainty in the ionization chamber was 2.43 times higher compared to the unshielded EFD diode at the smallest field size. **Conclusions:** Beam quality measurements, chamber selection procedure, and output factors were successfully carried out. A difference of up to 10% in output factor can occur if density scaling for electron density in virtual water is not considered. The uncertainty in output correction factors dominates, while positional and meter reading uncertainty makes a minor contribution to total uncertainty. An unshielded EFD diode is a preferred detector in small fields because of lower uncertainty in measurements compared to ionization chambers.

Keywords: Small field dosimetry, tomotherapy, TRS483, uncertainty

Introduction

In modern radiotherapy, small beamlet intensities impinging on the target from multiple directions accomplish the goal of implementing intensity-modulated radiation therapy (IMRT).[1] TomoTherapy is one such machine performing IMRT with a pioneering helical delivery system, integrated with computed tomography (CT) based image guidance.[2] It utilizes a dose calculation algorithm based on convolution-superposition.[3] Few studies[4,5] recommended the dosimetry in TomoTherapy based on commonly implemented code of practices (CoP) such as the American Association of Physicists in Medicine (AAPM) Task Group (TG) 51 protocol[6] and International Atomic Energy Agency (IAEA)-technical reports series (TRS) no. 398 protocol.[7] Moreover, to address the uniqueness of TomoTherapy hardware design, tailored quality assurance guidelines were reported by AAPM TG-148.[8]

The use of small beamlets in modern radiotherapy stresses to review the guidelines and procedures employed in standard dosimetric conditions for the following mentioned reasons:

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i. The primary challenge is the concept of field size.[9] Particularly, at small beam sizes, where the collimator shielding partially blocks photon source with considerable overlapping penumbrae, output cuts down at the central axis. This often breaks down the relation among the collimator setting (geometrical field size) and full width at half maximum (FWHM) of the lateral profile at the isocenter depth

ii. It is a challenge to ascertain the electron fluence in water and detector for small fields as it depends on beam quality. Determination of electron fluence lies on the foundation of charge particle equilibrium (CPE) condition, which is fulfilled in reference fields. However, in small fields, these approaches are usually inadequate due to loss of lateral CPE (LCPE), and instead, we count on Monte Carlo simulations

iii. The detector material, position, and size of the detector are determining components. Perturbation of the charged particle fluence, volume averaging, and the physical density of the detector's active volume perhaps intricate the application of ionization chamber for small field dosimetry. This infers that the existing perturbation factors in the IAEA TRS 398 or AAPM TG-51 may not be accurate for ionization to absorbed dose conversion. The beam quality correction factors available in TG-51 and TRS-398 has been established mainly for flattening filter beams. Other consequence results from the variation of beam spectrum with decreasing field size

iv. One of the critical steps in the treatment planning system (TPS) modeling is the correct experimental determination of output factors. It is a challenge to make accurate dose modeling in the TPS for small fields

To address the issues in small field dosimetry, through a collaborative work by AAPM and IAEA, an international CoP, TRS-483, was published in December 2017 for small static fields used in external beam radiotherapy.[9]

Many of the efforts in the past were invested in Monte-Carlo simulations of the TomoTherapy radiation characteristic and small field analytic models.[10,11] Howitz *et al*. determined both the experiment and Monte-carlo based beam quality index $TPR_{20,10}(10)$ at hypothetical 10 cm × 10 cm field size for the TomoTherapyHD machine.[12] Caprile *et al*. measured and calculated the output factors (OF) on the TomoTherapy machine with the smallest field size of 1.25 cm in the X-direction and fixed 5 cm jaw opening.[13] A recent publication by Lopes *et al*.[14] calculated the correction factor for the Standard Imaging A1SL ionization chamber. They also measured the beam quality index $TPR_{20,10}(10)$, and OF with the smallest field size of 1.25 cm in the X-direction. In our study, we accounted for the smallest field with a single leaf open field size of 0.625 cm in X-direction, and it also addresses an essential step of detectors selection for small field dosimetry. Output factors in both, water and virtual water phantom were determined to report the differences. Furthermore, a detailed investigation was performed to examine the positional and other factors of uncertainty in the measurement of output factors.

This study was taken as a part of a coordinated research project by IAEA with an agreement to measure output factors in various dosimetric conditions. Therefore, relative dosimetry of the small fields using the multi-leaf collimator (MLCs) and jaws of the TomoTherapy machine was studied and presented a detailed discussion on challenges faced following the recommendation of TRS 483. The flowchart in Figure 1 depicts the crucial step-wise procedures followed for beam quality and detector selection. This study aims to calculate the OF using different detectors and dosimetric conditions. It also estimates the various components of uncertainty in OF determination with different detectors.

Materials and Methods

TomoTherapy® Hi•Art® System (Accuray, USA) with a nominal 6 MV flattening filter-free (FFF) energy was used for the dosimetry. In the International Electrotechnical Commission coordinates system, the beam in Y-direction is collimated by adjustable jaws with an extension of 1.0 cm, 2.5 cm, and 5.0 cm at isocenter. Furthermore, 64 binary pneumatically driven tungsten leaves (32 on each side) modulate the beam to form small beamlets in X-direction.

Photon beam quality index $TPR_{20,10}(10)$

Experimental point data measurements were performed at a source to axis distance (SAD) of 85 cm in Scanditronix Wellhofer Gmbh I'mRT verification tool virtual water Phantom (RW3, density 1.045 g/cm³, the full size of the phantom is 33 cm × 36 cm × 18 cm) with some in-house customization [Figure 2]. Detectors used were two ionization chambers: IBA CC01 (IBA Dosimetry, Schwarzenbruck, Germany); PTW PinPoint 31006 (PTW-Freiburg, Germany) with a biasing voltage of + 300 Volts and one solid-state silicon IBA unshielded electron field diode (EFD) 3G with no bias voltage required. The ionization chamber central axis and diode sensitive volume (chip) were oriented perpendicular to the beam axis. In this orientation, the axis of symmetry of an ionization chamber and diode will be horizontal and vertical, respectively.

Point measurements were performed with the TomoTherapy electrometer (8 channel). The reference point of each detector was positioned in the beam central axis, at a depth of 10 cm and 20 cm, respectively, for a set of readings. Profile in Y-direction was acquired for accurate detector positioning by applying the beam center shift and MV image for visual verification. A set of readings was measured for 5 cm × 5 cm and machine-specific reference (msr) field 5 cm × 10 cm. Tissue phantom ratio $TPR_{20,10}(S)$, was calculated as a ratio of measurements at 20 cm and 10 cm depth.

The beam quality of TomoTherapy at the hypothetical reference field of 10 cm × 10 cm, $TPR_{20,10}(10)$, was determined using the analytical expression described by Palmans.[15]
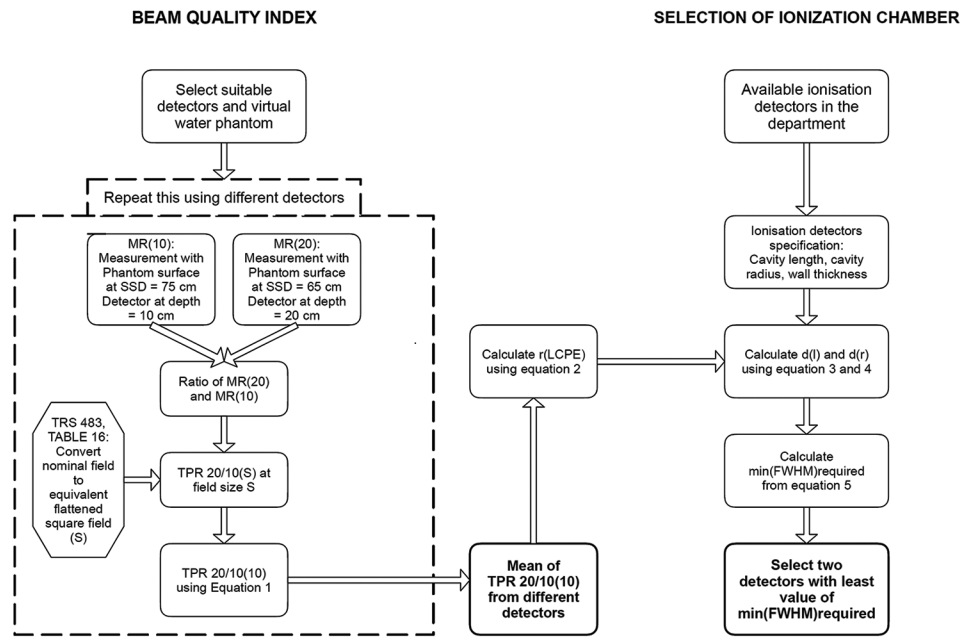
BEAM QUALITY INDEX

SELECTION OF IONIZATION CHAMBER

Figure 1: Flowchart describing the procedure adopted for beam quality measurement and ionization chamber selection

Figure 2: Setup of detector in a virtual water phantom for the measurement of beam quality (a) Ionization chamber (b) side view for the diode (c) bottom of the phantom for the diode

$$\mathbf{TPR}_{20,10}(10) = \frac{\mathbf{TPR}_{20,10}(\mathbf{S}) + \mathbf{c}(10-\mathbf{S})}{1 + \mathbf{c}(10-\mathbf{S})} \qquad (1)$$

Where, S: Equivalent flattened square field in cm, valid for $4 \leq S \leq 12$, Taken from TRS 483 and $c = (16.15 \pm 0.12) \times 10^{-3}$.

Selection of ionization chamber

As an ideal detector for small field does not exist, measurements performed with two or more type of detectors increase confidence. As per the IAEA guidelines, minimum of two ionization chambers and one diode (unshielded) for which the corrections factors are available in the TRS483 should be used in any study. In our study, one diode and two ionization chambers were used for relative dosimetry. From the available detectors in the department, the selection of the ionization chamber was based on the required minimum FWHM (min(FWHM)$_{required}$) of the small field that can satisfy the LCPE condition. At this condition, the external boundary of the detector volume in any direction must be at least r_{LCPE} away

from the field edges. The r_{LCPE} is a key factor which depends on the beam quality; the relation is given by:

$$\mathbf{r}_{LCPE} = 8.369 \times \mathbf{TPR}_{20,10}(10) - 4.382 \qquad (2)$$

where r_{LCPE} is in cm, and $TPR_{20,10}(10)$ value is determined from Equation 1.

If a detector has dimensions of the cavity length l, cavity radius r and a wall thickness t_{wall} (all dimensions in mm), then the outer boundaries of the detector in the longitudinal and radial direction, respectively, will be:

$$\mathbf{d}_l = \mathbf{l} + \mathbf{t}_{wall} \qquad (3)$$

$$\mathbf{d}_r = 2(\mathbf{r} + \mathbf{t}_{wall}) \qquad (4)$$

The condition of LCPE at the detector placed in the beam axis will be satisfied if the FWHM of the field fulfils the following condition:

$$\mathbf{min(FWHM)}_{required} \geq 2\mathbf{r}_{LCPE} + \mathbf{max}(\mathbf{d}_l, \mathbf{d}_r) \qquad (5)$$

The FWHM requirement condition was calculated for six ionization chambers (details given in the result section). Two ionization chamber with the lowest value of min(FWHM)$_{required}$ were selected for the dosimetry. Instead of an energy compensated shielded diode (photon diode), a better-suited unshielded EFD was chosen for small fields.

Determination of field output factors

The TomoTherapy electrometer measurement system application linked to both TomoTherapy water tank and TomoTherapy Electrometer (8-channel) was used for profile acquisition and analysis. The FWHM of a field is the most representative and essential field size parameter, which is used in selecting field output correction factors (perturbation factors). Beam profiles at 10 cm depth and 85 cm source to surface distance (SSD) were acquired with an EFD diode in TomoTherapy water phantom [Figure 3].

The TomoTherapy water tank mechanism allows to acquire a profile parallel to its length, a tank 90° rotation is required to acquire beam profile in another direction. Relative dosimetry was performed for clinical field sizes (f$_{clin}$) of 1 cm × 0.625 cm, 1 cm × 1.25 cm, 2.5 cm × 2.5 cm, 5 cm × 5 cm and 5 cm × 10 cm. Profiles were acquired with a step size ranging from 0.1 mm to 0.5 mm for the smallest and largest field size, respectively. The dosimetric field width in Y-direction (A) and X-direction (B) were analyzed at 50% of the profile maximum (FWHM) as recommended in TRS 483. The equivalent square field size, S$_{clin}$ is defined as:

$$S_{clin} = \sqrt{A \, X \, B} \qquad (6)$$

Dosimetric field widths in the SSD setup were multiplied with a geometrical scale factor of 85/95 = 0.8947 to get the field widths in the SAD setup.

Field output factor measurements were carried out at 10 cm depth with IBA CC01, PTW PinPoint ionization chambers and IBA EFD unshielded diode. Measurements in a water phantom were performed with 85 cm SSD. Measurements in a virtual water phantom were performed both in 85 cm SAD and SSD setup. Initially, for each field size, the beam profiles were acquired and the observed beam central axis shifts were applied to the chamber. Then the chamber was moved with small step-wise known shifts (~0.2 mm) in both lateral and longitudinal direction using a water tank motor system and couch, respectively, to acquire
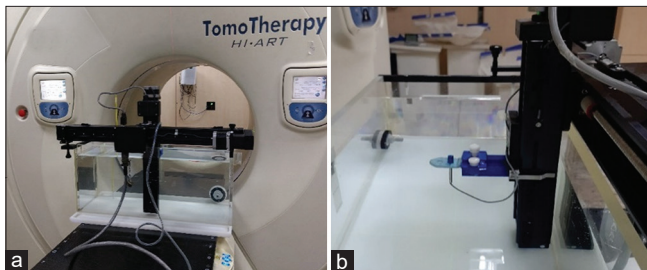

Figure 3: TomoTherapy water phantom (a) Setup with length parallel to X-direction (b) IBA unshielded electron field diode active volume/chip positioned perpendicular to the beam

the maximum output position. This step ensured the accurate positioning of the chamber in the beam central axis. The uncorrected output factor (UOF) is the ratio of corrected meter reading in a given clinical and msr field size. The OF was obtained by:

$$\Omega_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}} = \frac{M_{Q_{clin}}^{f_{clin}}}{M_{Q_{msr}}^{f_{msr}}} k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}} \qquad (7)$$

where, $M_{(Q_{clin})}^{(f_{clin})}$ and $M_{Q_{msr}}^{f_{msr}}$ are the corrected meter readings in the clinical field size and msr field size, respectively. $k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ is called the output correction factor, which takes into account the variation in beam quality between given two field sizes and volume averaging effect, valid at 10 cm depth. Consider both f$_{clin}$ and f$_{msr}$ as the S$_{clin}$ of clinical field size and msr field size, respectively.

The mass energy absorption coefficients of silicon are higher than water at low energies. Consequently, silicon diode detectors over-respond to low energy photons.[16] The unshielded diodes over-respond in large fields and underestimate the field output factors when they are normalized to large field size. An intermediate field method uses two detectors to limit energy dependence, and the OF are calculated as follow:

$$\Omega_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}} = \left[\frac{M_{Q_{clin}}^{f_{clin}}}{M_{Q_{int}}^{f_{int}}} k_{Q_{clin},Q_{int}}^{f_{clin},f_{int}}\right]_{det} \left[\frac{M_{Q_{int}}^{f_{int}}}{M_{Q_{msr}}^{f_{msr}}} k_{Q_{int},Q_{msr}}^{f_{int},f_{msr}}\right]_{IC} \qquad (8)$$

where, det and IC stand for suitable small field detector (unshielded EFD diode) and ionization chamber (CC01 ion chamber), respectively. The smallest field size without small field conditions has to be selected as an intermediate field size (int) for an ionization chamber. The factor $k_{Q_{int},Q_{msr}}^{f_{int},f_{msr}}$ can be considered as unity for the ionization chamber at a larger field size where small field conditions are absent and ion chamber energy dependence is minimum. The $k_{Q_{clin},Q_{int}}^{f_{clin},f_{int}}$ for the small field detectors were calculated using the equation:

$$\left[k_{Q_{clin},Q_{int}}^{f_{clin},f_{int}}\right]_{det} = \frac{\left[k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}\right]_{det}}{\left[k_{Q_{int},Q_{msr}}^{f_{int},f_{msr}}\right]_{det}} \qquad (9)$$

Output correction factors values as a function of S$_{clin}$, normalized at msr field, are given for a range of detectors in TRS 483.[9]

Uncertainty

The OF calculated from equation 7 has various components of uncertainty. Total uncertainty was calculated as per the methodology of Tolabin *et al.*[17]

$$U_{\Omega_{clin}}^2 = U_{M_{clin}}^2 + U_{M_{msr}}^2 + U_{k_{clin}}^2 \qquad (10)$$

where, $U_{k_{clin}}$ has a type B uncertainty which is estimated by means other than statistical treatment. Depending on the detector type and S$_{clin}$ of the field size, the value of uncertainty in the output correction factor ($U_{k_{clin}}$) can be taken from TRS 483.

The uncertainty in meter reading ($U_{M_{clin}}$ or $U_{M_{msr}}$) can be calculated as per equation 11 given below, incorporating positional or detector alignment uncertainty (U_{pos}) and uncertainty in electrometer reading relative to beam monitor (U_{rdg}).

$$U_{M_{clin\ or\ msr}}^2 = U_{pos}^2 + U_{rdg}^2 \qquad (11)$$

Evaluate the standard deviation (SD) of a set of electrometer readings for a particular f_{msr} to calculate the U_{rdgs}. The U_{pos} accounts for the uncertainties associated with the collimator setting (jaws and MLC), determination of beam center and accuracy of the scanning system. The equation can calculate it as:

$$U_{pos}^2 = U_{coll}^2 + U_{centre}^2 + U_{scan}^2 \qquad (12)$$

For all the field sizes at a depth of 10 cm and 85 cm SSD, acquire approximately 1 cm length X-dir and Y-dir beam profiles around the beam central axis with a step size of 0.1 mm. Then U = U_{coll} or U_{center} or U_{scan} can be calculated according to analytical formalism based on dose profile measurement by Lechner *et al.*[18]

$$U = \sqrt{\frac{Var(D(x))}{E(D(x))^2} + \frac{Var(D(y))}{E(D(y))^2}} \qquad (13)$$

Two independent second-order polynomials functions were fitted to two independent profiles (consisting of central 20 points normalized to D_{max}) in X and Y direction to express D(x) and D(y). Furthermore, two independent rectangular probability density functions (PDF) in X and Y direction for the detector position relative to maximum dose were considered. The expectation value (E) describing the average value and variance (Var) as the spread around the expectation for the dose profile can be computed from D(x or y) and PDF.

RESULTS

Photon beam quality index TPR$_{20,10}$(10)
The measurements of the beam quality TPR$_{20,10}$(S) are reported in Table 1. The maximum variation of TPR$_{20,10}$(S) at msr field size among the three detectors was 0.34%. We considered the field sizes of 5 cm × 10 cm and 5 cm × 5 cm which satisfied the criteria of $4 \leq S \leq 12$ for the calculation of TPR$_{20,10}$(10). A mean value of 0.627 was obtained, with a maximum deviation and SD of 0.32% and 0.001, respectively, representing the variation between detectors used.

Selection of ionization chamber
The minimum FWHM required for six detectors are reported in Table 2 along with the chamber specifications. The mean value of beam quality TPR$_{20,10}$(10) obtained in the previous section was used in equation 2, resulted in an r_{LCPE} value of 0.857 cm.

The IBA CC01 and PTW PinPoint chamber resulted in the least values of the min(FWHM)$_{required}$ for lateral charge particle

Table 1: Beam quality TPR$_{20,10}$(10) at hypothetical field size 10 cm × 10 cm calculated from the measured TPR$_{20,10}$(S) in a virtual water phantom using three detectors, where S is the equivalent flattened square field size taken from Table 16 of TRS 483 for a given field size

Measured beam quality, TPR$_{20,10}$(S)			
Field size (cm²)	PTW PinPoint	IBA CC01	EFD diode	Mean±SD
5×10	0.607	0.607	0.605	0.606±0.001
5×5	0.597	0.596	0.596	0.596±0.001
Calculated beam quality, TPR$_{20,10}$(10)			
S (cm)	PTW PinPoint	IBA CC01	EFD diode	Mean±SD
6.60	0.627	0.627	0.626	0.627±0.001
4.90	0.628	0.627	0.627	0.627±0.001

EFD: Electron field diode, SD: Standard deviation

equilibrium. Therefore, these two ionization chambers were selected for the field output factor determination, and a field size of 2.5 cm × 2.5 cm was taken as an intermediate field in this study.

Determination of field output factors
The measured field widths at SSD = 85 cm and depth = 10 cm are given in Table 3, along with the calculated equivalent field size in a SAD setup. The UOF for field sizes ≥2.5 cm × 2.5 cm and <2.5 cm × 2.5 cm showed a mean (SD) absolute variation of 0.002 (0.002) and 0.056 (0.031), respectively, among the three detectors. Maximum variation of UOF between CC01 chamber and EFD diode at the smallest field size was 3.24%, 5.10%, and 14.61% in SSD water, SSD slab, and SAD slab, respectively.

Measured UOF when multiplied with an output correction factor corresponding to equivalent field width (S_{clin}) at a depth of measurement, gives the OF. The OF for field sizes ≥2.5 cm × 2.5 cm and < 2.5 cm × 2.5 cm showed a mean (SD) absolute variation of 0.003 (0.003) and 0.046 (0.031), respectively, among the three detectors. Figure 4 illustrates field output factors in various setups of two ionization chambers and an unshielded EFD diode as a function of equivalent field size (S_{clin}). Furthermore, a comparison is shown for final field output factors calculated using the intermediate field method in water and virtual water (slabs). The maximum variation of OF between CC01 chamber and EFD diode at the smallest field size was 0.97%, 2.65%, and 11.80% in SSD water, SSD slab, and SAD slab, respectively. The deviation in OF between SSD water and SSD slab was up to 9.68% and 8.13%, respectively, for CC01 and EFD diode.

Intermediate field method was used for the field size smaller than 2.5 cm × 2.5 cm. For larger fields, the IBA CC01 chamber OF was considered because the ionization chamber has lower energy dependence. The final OF in SSD water and virtual water setup are reported in Table 4.
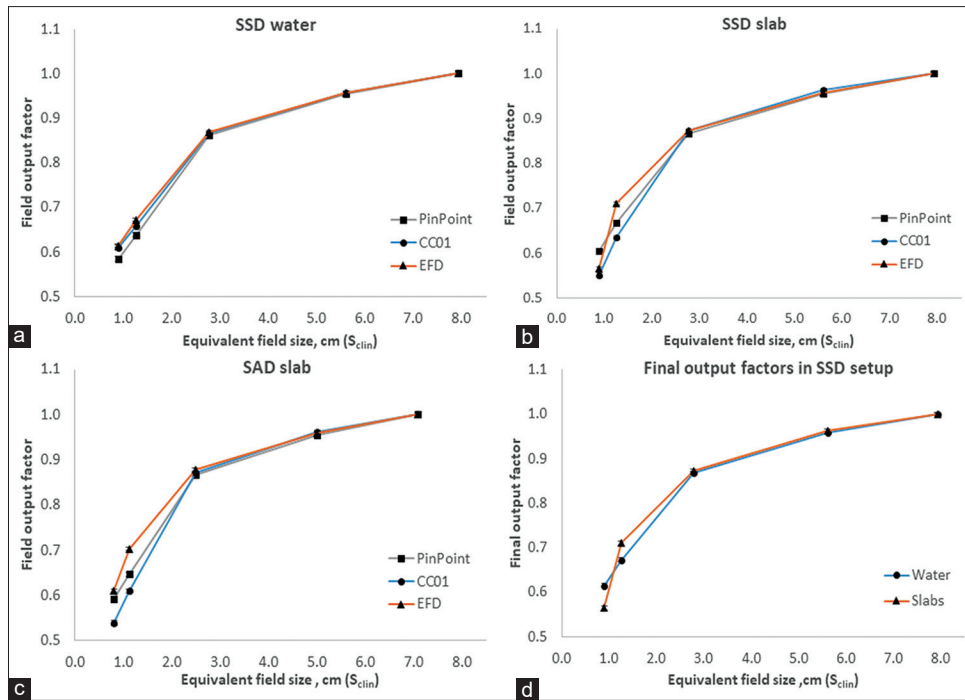
Figure 4: Field output factors of two ionization chambers and an unshielded EFD diode as a function of equivalent field size (S_{clin}) in various setups (a) Source to surface in water (b) Source to surface in slabs (c) Source to axis of detector in slabs. (d) A comparison is shown for final field output factors calculated using the intermediate field method in water and virtual water (slabs)

Table 2: The specifications of different ionization chambers along with the calculated minimum field width at half maximum required, minimum (full width at half maximum)$_{required}$, to satisfy the lateral charge particle equilibrium

Chamber	Cavity length, l (cm)	Cavity radius, r (cm)	Wall thickness, t_{wall} (g/cm²)	Density, ρ (g/cm³)	d_l (cm)	d_r (cm)	Calculated min(FWHM)$_{required}$ from r_{LCPE}
IBA FC-65G	2.310	0.310	0.073	2.250	2.342	0.685	4.073
PTW Semiflex	0.650	0.280	0.078	1.180	0.716	0.692	2.447
IBA CC13	0.580	0.300	0.070	1.760	0.620	0.679	2.410
IBA CC01	0.360	0.100	0.088	1.760	0.410	0.300	2.141
PTW PinPoint	0.500	0.100	0.085	1.180	0.572	0.344	2.303
SI A1SL	0.570	0.210	0.176	1.760	0.670	0.620	2.401

d_l and d_r: Outer boundaries of the ionization detector in longitudinal and radial direction, min(FWHM)$_{required}$: Minimum full width at half maximum, r_{LCPE}: Range of lateral charge particle equilibrium

Table 3: Calculated equivalent field size (S_{clin}) in source to surface distance and source to axis distance setups for different nominal field sizes, measured with IBA unshielded electron field diode

Nominal field size (cm²)	Measured field widths (SSD setup, 10 cm depth)		SSD setup, S_{clin}	SAD setup, S_{clin}
	Y (cm)	X (cm)		
5×10	5.65	11.13	7.93	7.10
5×5	5.65	5.56	5.61	5.02
2.5×2.5	2.80	2.76	2.78	2.49
1×1.25	1.15	1.37	1.26	1.12
1×0.625	1.15	0.70	0.90	0.80

SSD: Source to surface distance, SAD: Source to axis distance

Uncertainty

The positional uncertainty in the field output factor for the three detectors is summarized in Figure 5. At the smallest field size, PTW PinPoint ionization chamber has the highest uncertainty (U_{pos}) of 0.012%, whereas the IBA EFD diode has the least value of 0.009%. The positional uncertainty drops sharply with an increase in field size.

The uncertainty in meter reading for the three detectors is summarized in Figure 6. At the smallest field size, the unshielded EFD diode has uncertainty in meter reading (U_M) of 0.071% compared to < 0.02% shown by ionization chambers. A diode has a higher uncertainty in meter reading of 0.043% compared to ionization chambers even at msr field size.
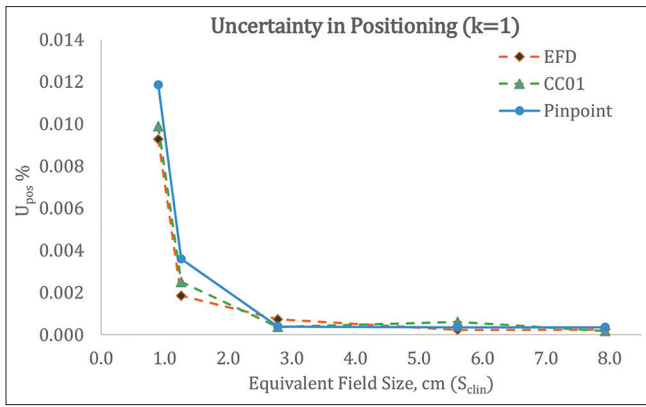
Figure 5: Comparison of unshielded EFD diode, PinPoint and CC01 ionization chamber for positional uncertainty in output factor with respect to equivalent field size
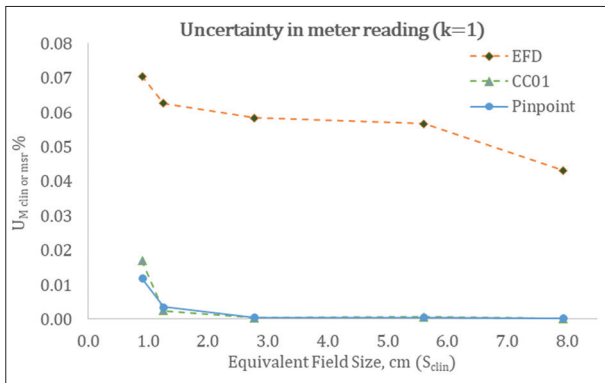

Figure 6: Comparison of unshielded EFD diode, PinPoint and CC01 ionization chamber for a meter reading uncertainty in output factor with respect to equivalent field size
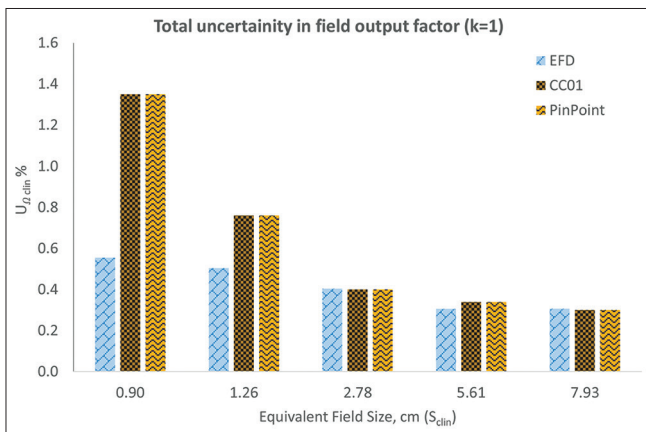

Figure 7: Total uncertainty in field output factor for IBA EFD unshielded diode, PTW PinPoint, and IBA CC01 ionization chamber

Due to higher uncertainty in the output correction factors of the ionization chamber, total uncertainty (U_{tot}) in the field output factors is 2.43 times higher in ionization chamber relative to EFD diode at the smallest field size as shown in Figure 7.

DISCUSSION

The publication of IAEA TRS 483 opened the gateway for testing small field dosimetry in more depth. Still, it comes with few challenges and limitations at the implementation side on various unconventional machines such as TomoTherapy.

The 85 cm bore diameter introduces some constraints in the use of the Accuray 2-D TomoTherapy water tank. The SAD setup with a chamber at 10 cm depth cannot be achieved with TomoTherapy water tank. Alternatively, if possible, the 3-D water tanks from PTW MP3-T Water Phantom System and IBA Blue Phantom Helix can be used. The CoP recommends 100 cm or as close as possible SAD for beam quality determination. Therefore, we performed the beam quality measurements at 85 cm SAD set up in the virtual water phantom against the recommended water phantom. Percentage depth dose was not taken as a beam quality specifier in our study because it required additional corrections if SSD was not equal to 100 cm. Determination of these additional correction factors requires tissue maximum ratio and normalized peak-scatter factor for the field size S at 85 cm SSD,[19] which are not available. As reported by Howitz *et al*.,[12] the beam quality calculated by Monte-Carlo simulation was 0.628 and 0.631 from two different calculation methods and the experimental mean value of 0.635, all agreed within 1.28% to our measured value of 0.627.

The low-energy photons from the collimator and phantom scatter are minimal in small fields. In the absence of LCPE, the low-energy photon contribution to the field center is further reduced. The PTW PinPoint 31006 and PTW PinPoint 31014 (new model of PTW PinPoint 31006) differed only at lower energies.[20] Thus, even though the OF measurement was done with PTW PinPoint 31006, the field output correction factor of PTW PinPoint 31014 provided in the TRS 483 was used.

The profile measurements were done in a water tank with EFD diode at 85 cm SSD and 10 cm depth. The 90° water tank rotation is necessary to acquire longitudinal and lateral profiles. According to Accuray protocol, field width in X-direction is calculated at 25% of the dose maximum.[21] This protocol, in general, applies to larger field sizes where the shape of the profile in X-direction is similar to conventional FFF beams in a linac. However, at small field sizes, the shape of profiles is not like conventional FFF beams. TRS 483 recommends both field widths in X and Y directions to be measured at 50% of dose maximum (FWHM) to calculated S_{clin}. The beam central axis in TomoTherapy passes through a plane between two central MLCs. Therefore, the smallest field size of 1 cm × 0.625 cm is an off-axis field with a single open MLC. A shift (~3 mm) calculated from profile needs to be applied to the chamber for accurate field output factor measurement. The measured field widths in Y-direction (collimated by jaws) were within the tolerance limit of 1% as per the TG-148[8] from the baseline values during commissioning.

The UOF for different detectors was calculated as a ratio of meter reading for a clinical field size and msr field size. The temperature, pressure, and humidity were nearly constant during the measurement. The measured ion recombination and polarity correction factor for all field sizes was <0.1%. The overall effect of these correction factors in a corrected meter reading will be negligible; additionally, these correction factors will cancel out in the ratio of corrected meter readings. Therefore, these terms can be accounted for in the uncertainty of meter reading. For calculating the OF, values of the output correction factors were taken from TRS-483. The detector specific output correction factor increase with an increase in deviation from the field size needed to satisfy LCPE ($\min(\text{FWHM})_{required}$) at the detector. Therefore, select the detectors with the minimum value of the $\min(\text{FWHM})_{required}$ which also reduces the volume averaging effect. The maximum difference between OF and UOF was 0.92%, 0.58%, 0.35% for EFD diode, CC01 and PinPoint chambers respectively at intermediate field size of 2.5 cm × 2.5 cm. At the smallest field size, the maximum difference between OF and UOF was 0.99% and 2.23% for the EFD diode and CC01 chamber. The PinPoint OF at 1 cm × 0.625 cm was not considered for comparison with CC01 and EFD because the output correction factor was extrapolated below 1.2 cm field size and the corresponding value in Table 4 is the UOF multiplied with an extrapolated output correction factor.

TRS 483 recommends water as a reference medium for measurements, which requires extra care and effort to handle water phantoms. Virtual water phantoms (slabs) are growing in popularity for their easiness. On the downside, slabs have fixed thicknesses which restricts the variability in the depth of measurement. Therefore, depth scaling according to the ratio of electron densities in water and slabs was not done; instead, the OF measurements were performed at a fixed physical depth of 10 cm as per the IAEA agenda. This led to a maximum discrepancy of 4.7%, 8.13%, and 9.69% between SSD water and SSD slab with Pinpoint, EFD diode, and CC01 chamber, respectively.

Beam modeling and commissioning procedures in are performed by the manufacturers.TomoTherapy. Beam data provided by users for the TPS modeling is not significantly focused on the small field measurements. Many challenges

prevail both in small field measurements as well as calculations. With accomplishments in the measurement accuracy of small fields, the accuracy of source models can be tested. It needs the willingness of the manufacturers to optimize their source models for small fields. Moreover, these small fields OF measurements can be included in the machine acceptance protocols and testing in future. At the smallest field size, the UOF in different setups and detectors has 3.28% SD, which dropped to 2.19% for OF. Mamesa *et al.* in their study reported that when corrected field output factors are used instead of uncorrected field output factors, then it reduces the discrepancy in calculated monitor units from beam commissioning datasets in Eclipse™ TPS using various chambers.[22] The evaluation of such discrepancy for TomoTherapy is not possible at present because of the limited options available in TPS. The verification of OFs by recalculation and comparison against measurements can be a part of the quality assurance of TPS.

A small field comes with associated uncertainty in measurement. Specifically, for TomoTherapy, there is a lack of proper uncertainty estimation in the various steps involved in the determination of output factors. This study presents a detailed calculation of uncertainty for ionization chambers and diode. The positional uncertainty, as proposed by Lechner *et al.*,[18] along with the overall methodology by Tolabin *et al.*,[17] was used to calculate total uncertainty in output factors. As expected, the positional uncertainty in both the diode and ionization chamber reduces with an increase in field size. A diode detector has a lower positional uncertainty compared to the ionization chamber in small fields. In contrast, the EFD diode has a higher uncertainty in meter reading for all field sizes compared to the ionization chamber. A significant contribution to total uncertainty came from the uncertainty in output correction factors of the detector. This resulted in a lower total uncertainty for EFD diode in comparison to the ionization chambers. Prior to selection of any detector, it is suggested to refer of the TRS 483 to get an idea about the detector uncertainty range and to reduce the overall uncertainty in output factors. The field size of 1 cm × 0.625 cm at 10 cm depth in SSD setup has an FWHM of (A) 1.15 cm in Y-direction and (B) 0.70 cm in X-direction with an aspect ratio of 1.64. A larger uncertainty at the smallest field size can be considered in Figure 7 for violation of the aspect ratio 0.7< A/B <1.4 conditions. We have considered the output correction factor uncertainty the same as provided of the TRS 483.

The leakage for the used detectors was estimated for at least 5 min before its actual irradiation for the said purpose and the observed leakage per minute was negligible. As per TRS 483, the leakage should be smaller than 0.1% of the detector reading. The estimated Type A absolute uncertainty in meter reading with these detectors was 0.000431.

CONCLUSIONS

The relative dosimetry in TomoTherapy for small fields was carried out following the recommendations of IAEA TRS 483.

Table 4: Final field output factors in water and virtual water (slabs) with an intermediate field method

Nominal field size (cm^2)	Equivalent field size S_{clin} (cm)	$\Omega^{fclin, fmsr}_{Qclin, Qmsr}$ Water	$\Omega^{fclin, fmsr}_{Qclin, Qmsr}$ Virtual water
5×10	7.930	1.000	1.000
5×5	5.606	0.958	0.964
2.5×2.5	2.778	0.868	0.872
1×1.25	1.256	0.673	0.711
1×0.625	0.895	0.614	0.565

Setup was done with a SSD of 85 cm and depth 10 cm using IBA unshielded EFD diode and IBA CC01 ionization chamber. SSD: Source to surface distance, EFD: Electron field diode

The $\mathrm{TPR}_{20,10}(S)$ satisfying the condition of 4 cm ≤ S ≤ 12 cm was used to calculate $\mathrm{TPR}_{20,10}(10)$, resulting in a beam quality of 0.627 ± 0.001 at hypothetical field size of 10 cm × 10 cm. Ionization detectors were selected for the relative dosimetry based on the minimum FWHM required for the existence of LCPE. The measurement of field output factor at small fields was successfully carried out both in water and virtual water phantom. A difference of up to 10% can occur if density scaling for electron density in virtual water is not considered. Ionization chamber and unshielded EFD diode results are similar for field output factors at field size ≥2.5 cm × 2.5 cm. Positional and meter reading uncertainty make a minor contribution to the total uncertainties of output factors. The uncertainty in output correction factors dominates all other uncertainty, and the total uncertainty in the ionization chamber is 2.43 times higher compared to unshielded EFD diode at the smallest field size. An unshielded EFD diode is a preferred detector for field output factor measurement at field sizes smaller than 2.5 cm × 2.5 cm due to its lower total uncertainty.

Conflicts of interest
There are no conflicts of interest.

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