Iranian Journal of Medical Physics

ijmp.mums.ac.ir

A Study of the Impact of TRS-483 Recommended Correction Factors for Dosimetry of Small Fields in Flattening Filter Free Beams used in TrueBeam Linear Accelerators

Gopinath Mamballikalam[1,3*], S.Senthil Kumar[2], C.O.Clinto[3], Jaon bos R.C. [3], Ahamed Basith P.M.[3], Ganesh K.M.[4]

1. R&D, Bharathiar University, Coimbatore, Tamilnadu, India
2. Rajaji Hospital & Madurai Medical College, Madurai, Tamilnadu, India
3. Aster Medcity, Kochi, Kerala, India
4. Kidwai Memorial Institute of Oncology: Bangalore, Karnataka, IN

ARTICLE INFO

Article type:
Original Paper

Article history:
Received: Feb 01, 2020
Accepted: Jun 23, 2020

Keywords:
FFF Beams
TRS-483
Linear Accelerator
Small Field Dosimetry

ABSTRACT

Introduction: To study the effect of the International Atomic Energy Agency (IAEA) TRS-483 recommended beam quality correction factor in reference dosimetry and to examine the recommended field output correction factor for relative dosimetry of 6-MV flattening filter free (FFF) small fields, used in a Varian TrueBeam linear accelerator (LINAC).

Material and Methods: The beam quality and field output correction for 6-MV FFF beams were adopted from the TRS-483 protocol. Monte Carlo (MC) simulation of the output factor was performed using the PENELOPE-based PRIMO software and compared with the TRS-483 corrected output factors. Two analytical anisotropic algorithm (AAA) models in the Eclipse™ treatment planning system (TPS) were created; one with an output factor taken as the ratio of meter readings and one with an output factor obtained by multiplying the TRS-483 correction factor by the ratio of meter readings. Besides, box field and dynamic conformal arc (DCA) plans were created for both AAA models for verification and validation. The patient-specific quality assurances (QA) for ten different targets were performed, and deviations between the measured and TPS-calculated point doses in both models were examined.

Results: Separate beam quality correction factors for FFF beams in the TRS-483 protocol only resulted in an improvement of 0.1% in reference dosimetry. The TRS-483 corrected output factor was in a better agreement with the MC-calculated output factor. For a patient-specific QA of DCA plans, the output factor-corrected AAA dose calculation algorithm showed a better agreement between the measured and simulated doses. Also, there was a smaller deviation (1.2%) for the smallest target of 0.23 cc (8 mm equivalent sphere diameter) used in this study.

Conclusion: The field output factors for the LINAC small beams can be improved by incorporating the TRS-483 correction factors. However, the extent of improvement that can be expected depends on the source model of the calculation algorithm and how these well-generalized corrections are suitable for user beams and detectors.

► *Please cite this article as:*
Mamballikalam G, Kumar SS, Clinto CO, Jaon bos RC, Ahamed Basith PM, Ganesh KM. A Study of the Impact of TRS-483 Recommended Correction Factors for Dosimetry of Small Fields in Flattening Filter Free Beams used in TrueBeam Linear Accelerators. Iran J Med Phys 2021; 18: 178-186. 10.22038/ijmp.2020.46236.1722.

Introduction

In recent years, technological advances have led to the increased use of small photon fields for intracranial and extracranial radiotherapy. Dosimetric accuracy, along with positioning accuracy, is critical for high-precision radiotherapy of small targets by stereotactic radiosurgery (SRS), stereotactic radiotherapy (SRT), and stereotactic body radiation therapy (SBRT). On the other hand, use of flattening filter free (FFF) beams has become very common in SRS and SRT, even in extracranial radiotherapy [1, 2], using conventional linear accelerators (LINACs); however, this has reduced their traceability to reference dosimetry and increased the uncertainty of clinical dosimetry. Also, dosimetry in non-reference small fields with non-standard unflattened photon beams still requires further research.

The availability of high-definition, 2.5-mm multi-leaf collimators (MLCs), a six-dimensional (6D) couch, kilovoltage cone-beam computed tomography (kV-CBCT), and an online BrainLAB ExacTrac® imaging system has made it possible to use conventional LINAC for high-precision radiotherapy [3]. Modern LINAC systems additionally provide FFF beams with higher dose rates, which significantly reduces the beam-on time. The IAEA TRS-398 and AAPM TG-51 protocols are widely adopted codes of practice (CoP)

*Corresponding Author: Tel: +91 9656900749; Email: mgnmenon@gmail.com

for the reference dosimetry of external beam radiotherapy [4, 5]. By applying these protocols, ionization chambers can be calibrated in terms of absorbed dose to water in a standard laboratory with a reference beam (Co-60) quality in a field size of 10×10 cm^2.

Standard laboratories provide calibration coefficients for users to perform reference dosimetry. Beam quality correction factors are only provided for flattened beams as a function of tissue phantom ratio ($TPR_{20,10}$) or percentage depth dose (PDD_{10}). The fluence of unflattened and flattened photon beams is dosimetrically different [6]. The treatment planning system (TPS) used for radiotherapy simulation is modeled based on the acquired beam parameters, such as PDD, profile, and output factors (OFs). Any deviation in these parameters may affect the simulation results; therefore, the user needs to feed proper beam data and validate the beam model [6].

The treatment plan verification results have shown more deviations for small target sizes as compared to larger targets. Disagreements between the simulated and measured doses for small targets encouraged us to perform further investigations on small field dosimetry [7]. The effect of small field uncertainty is significant in the OF determination, which directly influences the monitor unit (MU) calculations [8]. The measurement of OFs for small fields, by determining the ratio of detector readings between a small clinical field and a 10×10 cm^2 reference, is inaccurate [9]. Overall, selection of the detector and correction factors play an important role in the OF determination. It seems necessary to standardize the protocols to account for small field conditions as an extension of the existing CoP.

In this regard, Alfonso et al. proposed a new formalism for the dosimetry of small and non-standard fields in 2008 [10]. Pantelis E et al. [11] implemented this new formalism in the CyberKnife system. Besides, Ralston et al., in 2012, proposed a method for correcting the volume average effect [12]. By applying this correction, W. Lechner et al. [13] concluded that for the majority of detectors, the dose response ratio was within the measurement uncertainty range when irradiated with flattening filter (FF) and FFF beams. The lower density of air as compared to water is the cause of under-response, even after the volume-averaging correction for all air-filled ionization chambers. In 2014, Benmakhlouf et al. [14] conducted a PENELOPE Monte Carlo (MC) study to derive the detector-specific OFs for small fields. Azangwe et al. [15] also described detector-specific correction factors in an experimental study.

Three physical conditions are considered to designate a photon beam as a small field; this can result in an overlap between the field penumbrae and the volume of the detector [16-18]. Overall, loss of lateral charged-particle equilibrium on the beam axis and partial occlusion of the primary photon source by the collimating system are beam-related issues. On the other hand, a similar-sized or larger detector as compared to the beam dimensions can cause volume averaging and significant fluence perturbation, which are detector-related issues. This perturbation effect depends on the construction details of the ion chamber. The absorbed dose measurement is significantly affected by fluence perturbation for small fields, compared to broad beams [19, 20].

In 2017, the International Atomic Energy Agency (IAEA) in collaboration with the American Association of Physicists in Medicine (AAPM) published TRS-483 as a new CoP for small-field dosimetry [21]. This formalism described for small-field dosimetry was similar to that proposed by Alfonso et al. in 2008. The TRS-483 protocol provided correction factors for reference and relative dosimetry by considering the influential small-field parameters and compiling the published data from journals.

In this regard, M. Saiful Huq et al. [22] carried out a dosimetric evaluation of the new CoP for conventional LINAC beams and compared it with the existing protocols. There is a general understanding that TRS-483 is mostly related to the CyberKnife system and TomoTherapy. Therefore, the present study aimed to investigate the effect of TRS-483 CoP on the dosimetry of 6-MV FFF small photon fields, used in a Varian TrueBeam® LINAC for hypofractionated radiotherapy of small targets.

Materials and Methods

Overview of TRS-483 protocol

Reference dosimetry

The absorbed dose to water in a machine-specific reference (msr) field size for an FFF user beam in the absence of a chamber can be obtained by Eq. 1 [21]:

$$D^{f_{msr}}_{W,Q^{FFF}_{msr}} = M^{f_{msr}}_{Q^{FFF}_{msr}} . N^{f_{ref}}_{D,W,Q_0} . k^{f_{ref}}_{Q^{WFF},Q_0} . k^{f_{ref}}_{Q^{FFF},Q^{WFF}} . k^{f_{msr},f_{ref}}_{Q^{FFF}_{msr},Q^{FFF}} \quad (1)$$

where $k^{f_{ref}}_{Q^{WFF},Q_0}$ represents the beam quality correction factor from the calibrated Co-60 beam quality to conventional beams with flattening filter (WFF); same as k_{Q,Q_0} correction used in TRS-398; $k^{f_{ref}}_{Q^{FFF},Q^{WFF}}$ represents the difference in the ionization chamber response between WFF and FFF beams; and $k^{f_{msr},f_{ref}}_{Q^{FFF}_{msr},Q^{FFF}}$ denotes the quality difference between the reference 10×10 cm^2 field size and msr field size. The concept of msr field is only applicable for machines that cannot establish a conventional 10×10 cm^2 reference field. For a conventional LINAC, Eq. 1 was simplified to Eq. 2a and Eq. 2b [21]:

$$D^{f_{ref}}_{W,Q^{FFF}} = M^{f_{ref}}_{Q^{FFF}} . N^{f_{ref}}_{D,W,Q_0} .(k^{f_{ref}}_{Q^{WFF},Q_0} . k^{f_{ref}}_{Q^{FFF},Q^{WFF}}) \quad (2a)$$

$$D^{f_{ref}}_{W,Q^{FFF}} = M^{f_{ref}}_{Q^{FFF}} . N^{f_{ref}}_{D,W,Q_0} . k^{f_{ref}}_{Q^{FFF},Q_0} \quad (2b)$$

where $k^{f_{ref}}_{Q^{FFF},Q_0}$ is the required beam quality correction from the Co-60 reference beam to the 10×10 cm^2 FFF user beam. This value is obtained as a product of $k^{f_{ref}}_{Q^{WFF},Q_0}$, correction factors for the difference in water/air stopping power ratios, and volume-averaging correction factors. These correction factors, as a function of $TPR_{20,10}$ and PDD_{10}, are provided in TRS-483 for different chambers.

Relative dosimetry

The absorbed dose to water for a clinical field (f_{clin}) can be obtained from the reference dose of the msr field (f_{msr}) by multiplying it by the field OF in Eq. (3) [21]. The OFs are also called the total scatter factors [15, 23, 24] or relative dose factors:

$$D^{f_{clin}}_{W,Q_{clin}} = D^{f_{msr}}_{W,Q_{msr}} . \Omega^{f_{clin},f_{msr}}_{Q_{clin},Q_{msr}} \tag{3}$$

For machines with a 10×10 cm^2 reference field, f_{msr} can be replaced with f_{ref}. Therefore, $\Omega^{f_{clin},f_{ref}}_{Q_{clin},Q_{ref}}$ is the field OF that converts the absorbed dose to water for f_{ref} to the absorbed dose to water for f_{clin} (Eq. 4) [21]:

$$\Omega^{f_{clin},f_{ref}}_{Q_{clin},Q_{ref}} = \left(\frac{D^{f_{clin}}_{Q_{clin}}}{D^{f_{ref}}_{Q_{ref}}} \right) \tag{4}$$

The field OFs can be derived from the ratio of detector readings according to Eq. 5 [21]:

$$\Omega^{f_{clin},f_{ref}}_{Q_{clin},Q_{ref}} = \left(\frac{M^{f_{clin}}_{Q_{clin}}}{M^{f_{ref}}_{Q_{ref}}} \right) \times k^{f_{clin},f_{ref}}_{Q_{clin},Q_{ref}} \tag{5}$$

To obtain the field OF, the ratio of meter readings was multiplied by the correction factor. $k^{f_{clin},f_{ref}}_{Q_{clin},Q_{ref}}$ denotes the difference between the detector responses in f_{clin} and f_{ref}. For large fields, the correction factor is close to unity. However, for small fields smaller than 3×3 cm^2, the field output correction factors are predominant. For these small fields, there is no optimal detector, and if the used detector is very small and energy-independent, the required correction will be minimal. The TRS-483 protocol provides output correction factors as detector-specific generic values, which are a function of equivalent square field sizes from the MC calculations and experimental results.

The TRS-483 formalism also provides a method to obtain the field OF, using an intermediate field method (IFM). If a single detector for the entire range of field sizes is not obtainable, a field is defined as small as possible without small field conditions and called the intermediate field f_{int}. For fields larger than f_{int}, an ionization chamber can be used, and for smaller fields, a suitable small field detector can be applied. By IFM, OF can be obtained using Eq. 6a and Eq. 6b [21]:

$$\Omega^{f_{clin},f_{ref}}_{Q_{clin},Q_{ref}} = \left[\Omega^{f_{clin},f_{int}}_{Q_{clin},Q_{int}}\right]_{det}\left[\Omega^{f_{int},f_{ref}}_{Q_{int},Q_{ref}}\right]_{IC} \tag{6a}$$

$$\Omega^{f_{clin},f_{ref}}_{Q_{clin},Q_{ref}} = \left[\left(\frac{M^{f_{clin}}_{Q_{clin}}}{M^{f_{int}}_{Q_{int}}}\right) \times k^{f_{clin},f_{int}}_{Q_{clin},Q_{int}}\right]_{det}\left[\left(\frac{M^{f_{int}}_{Q_{int}}}{M^{f_{ref}}_{Q_{ref}}}\right) \times k^{f_{int},f_{ref}}_{Q_{int},Q_{ref}}\right]_{IC} \tag{6b}$$

Experimental setup and methods

All measurements were carried out in Varian TrueBeam® LINAC for 6-MV FFF beams. The $TPR_{20,10}$ value of the beams was 0.632. The dose rate for the measurements was set at a maximum of 1400 MU/min. All measurements were done in 100-cm SSD in a 10-cm depth, using an IBA Blue Phantom 2 radiation field analyzer and IBA Dose 1 electrometer. Reference dosimetry was performed in a 10×10 cm^2 field size, using an IBA FC65G 0.65-cc ionization chamber (IC). The beam quality correction for 6-MV FFF beams, provided in TRS-483, was used in reference dosimetry. Also, the K_Q value for the FFF beam could be obtained from the K_Q value of WFF beam by multiplying the correction factors to account for the difference in water-to-air stopping power ratio and volume averaging.

Field output measurements were performed using CC01 IC slots for equivalent square field sizes <3 cm and using CC13 for equivalent square field sizes of 3 cm or larger. The IBA CC01 is a 10-mm^3 mini ionization chamber, with a 0.5-mm C-552 wall and a 0.35-mm diameter steel electrode. Besides, the IBA CC13 is a standard IC of 150 mm^3 active volume, with a 0.4-mm C-552 wall and a 1-mm diameter C-552 central electrode. Both chambers were oriented with their stems perpendicular to the beam axis. For output measurements, meter readings were normalized to the 10×10 cm^2 reference field size. The field output correction factors were adopted from the TRS-483 protocol for CC01 and CC13 and plotted against the equivalent square field sizes in Figure 1. The CC01 was used for equivalent square field sizes <3 cm and CC13 for field sizes of 3 cm or larger. In field sizes <3 cm, CC01 required less correction as compared to CC13, while in field sizes >3 cm, CC13 did not require any corrections. For CC01, the correction factor was significant up to 8 cm, which might be due to the effect of fluence perturbation in the presence of steel electrode.

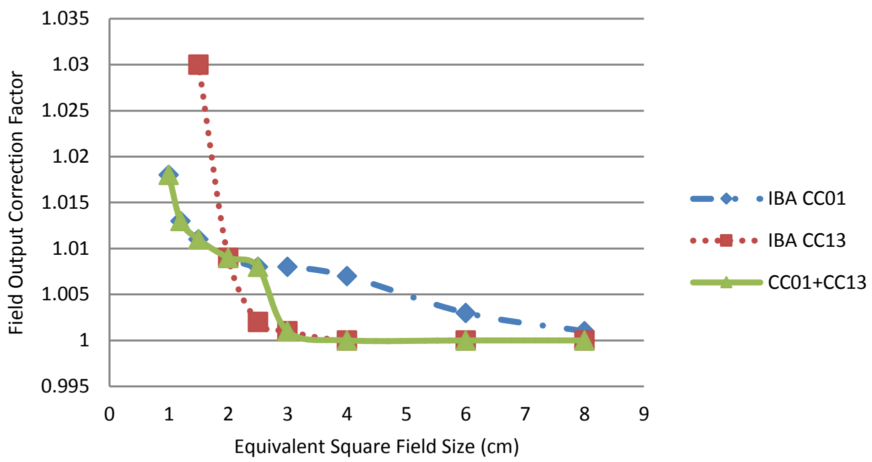

Figure 1. Field output correction factors versus equivalent square field sizes for the IBA CC01 and CC13 ionization chambers (ICs). The gray line shows the correction factors against the equivalent square field sizes for the combined use of CC01 and CC13 CIs.

The TRS-483 corrected field OFs were compared with the MC-calculated values. The MC simulation was performed using the PENELOPE-based PRIMO software. The phase-space files, generated by Varian VirtuaLinac application, were used as input sources for the OF simulation. In MC simulations, to simulate the clinical beam, the geometry of the accelerator must be known. Particle information, such as energy, position, and direction of flight, is included in the phase-space file. If the phase-space file contains large numbers of particles, it serves as the source of radiation, and the geometry of the accelerator can be neglected [25]. The first-generation phase-space files have been validated by Gete et al. [26], and the second-generation phase-space files have been validated by Belosi et al. [27]. Belosi et al. also stated that accurate radiation source simulation can be achieved using Varian phase-space files.

The machine parameters used to generate the phase-space files for 6-MV FFF beams were as follows: mean energy of the incident electron beam (energy E)=5.9 MeV; Gaussian energy spread, dE=0.051 MeV; Gaussian special spread in the 'X' direction (FWHM), Spot X (σ_x)=0.6645 mm; Gaussian special spread in the 'Y' direction (FWHM), Spot Y (σ_y)=0.7274 mm; and source beam divergence=0.0573°.

Generally, PRIMO is an MC-based LINAC simulator and a dose calculator. It was developed based on the PENELOPE 2011 code, PenEasy code, dose planning method, and PenEasy LINAC; it can also import and simulate other codes in the IAEA binary format. A graphic user interface allows the user to configure the simulated machine, create a phantom, or import computed tomography (CT) images; after simulation, the absorbed dose distribution can be seen in phantoms or on CT images. The simulation in PRIMO comprises of three segments: (1) beam parameters, which allow the user to define the initial electron energy, FWHM, and beam divergence; (2) field configuration, as the collimator setting can be defined; and (3) dose tallying, where the patient modeling can be done, the phantom can be created, or the CT scan can be imported.

In the Eclipse™ TPS, the AAA 13.6.17 dose calculation algorithm was used for beam modeling and virtual simulation [28]. For field sizes smaller than 1×1 cm^2, the Eclipse TPS does not use OFs for beam modeling. The OFs have three components: (1) phantom scatter, (2) head scatter, and (3) collimator backscatter into the monitor chamber. The collimator backscatter radiation represents the amount of radiation scatter from the collimator system into the monitor chamber. In the AAA photon beam source model, the collimator backscatter factor (CBSF) for a jaw-defined field (X, Y) was estimated from the measured OF table using Eq. 7 [28, 29]:

$$CBSF(X,Y)=\frac{OF_{ref}}{OF(X,Y)}.\frac{D^{'}(X,Y)}{D^{'}_{ref}} \quad (7)$$

where $\frac{OF_{ref}}{OF(X,Y)}$ is the ratio of measured OFs with respect to the reference field size; and is the ratio of absolute dose component calculated by the AAA model, based on the calibration calculations for the reference field size. The source model-calculated CBSF values were used in this study.

In this study, we developed two AAA models; one with an OF taken as the ratio of meter readings and one with an OF obtained by multiplying the TRS-483 correction factor by the ratio of meter readings. For validation, four field plans were created in the Eclipse TPS for both AAA models to deliver a 12-Gy dose for jaw-defined field sizes of 1×1 cm^2, 2×2 cm^2, 3×3 cm^2, 4×4 cm^2, 6×6 cm^2, 10×10 cm^2, and 15×15 cm^2. Point dose measurements were done in a RW3 slab phantom, with CC01 IC placed in a 10-cm depth; the measurements were compared with the TPS-calculated doses. The dynamic conformal arc (DCA) plans were also developed for ten patients with different target sizes and prescriptions in this study. Besides, a patient-specific QA was performed in the RW3 slab phantom using the CC01 IC. Deviations in point doses between the measured and simulated TPS values for both models were analyzed.

Results

Based on the present results, the $k_{Q_{FFF}}^{f_{ref}}$ value for the 6-MV FFF beam ($TPR_{20,10}$=0.632) was 0.997. It was obtained as a product of $k_{Q_{WFF}}^{f_{ref}}$ (0.996), based on the correction factor for the difference in water-to-air stopping power ratio (0.999) and volume averaging correction factor (1.002). There was no separate beam quality correction table for the FFF beam in TRS-398. According to the TRS-398 table, the $k_{Q}^{f_{ref}}$ value was 0.998. There was only an improvement of 0.1% in the beam quality correction for the FC65G 0.65-cc IC in the reference dosimetry in a field size of 10×10 cm^2.

The comparison of OFs obtained from the ratio of meter readings, TRS-483 corrected factors, and MC-simulated factors against the equivalent square field sizes is shown in Table 1, and variations are presented in Figure 2. The field OF, obtained as the ratio of meter readings, was much lower than the MC-simulated OF below the equivalent square field size of 2 cm; this is a clear indication of measurement inaccuracy in small fields. The TRS-483 corrected OFs were superior to those uncorrected with MC-simulated values. For a field size of 2 cm, the correction was sufficient, while a 1.8% correction for a field size of 1 cm was inadequate.

Table 1. Comparison of output factors (OFs) obtained from the ratio of meter readings, TRS-483 corrected OFs, and MC-simulated OFs against equivalent square field sizes

Equivalent square field size (cm)	Chamber	Measured OFs	Correction factor	TRS-483 corrected OFs	MC-simulated OFs
1	CC01	0.635	1.018	0.646	0.684
2	CC01	0.786	1.009	0.793	0.802
3	CC13	0.843	1.001	0.844	0.842
4	CC13	0.876	1.000	0.876	0.875
5	CC13	0.896	1.000	0.896	0.899
10	CC13	1.000	1.000	1.000	1.000

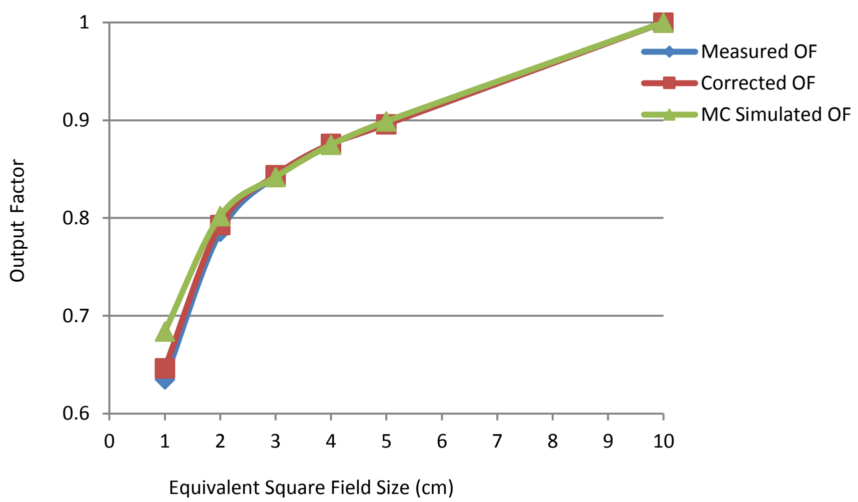

Figure 2. The output factors (OFs) versus the equivalent square field sizes obtained by measurements, TRS-483 correction, and MC simulation.

The OF and CBSF obtained with and without correction for the equivalent square field sizes are shown in Table 2. Overall, 1.8%, 0.9%, and 0.1% positive corrections for 1×1 cm^2, 2×2 cm^2, and 3×3 cm^2 jaw field sizes resulted in similar corrections for CBSF in negative correction. As the field size decreased, the OF also decreased, while the backscattered radiation from the collimator to the dose monitor chamber increased (Figure 3).

Table 2.The output factor (OF) and collimator backscatter factor (CBSF) obtained with and without correction for the equivalent square field sizes

Equivalent square field size (cm)	Output correction factor	Without correction		With correction		CBSF correction factor
		Measured OF	Calculated CBSF	OF	Calculated CBSF	
1	1.018	0.635	1.163	0.646	1.143	0.983
2	1.009	0.786	1.050	0.793	1.041	0.991
3	1.001	0.843	1.020	0.844	1.020	1.000
4	1.000	0.876	1.015	0.876	1.015	1.000
6	1.000	0.931	1.004	0.931	1.004	1.000
10	1.000	1.000	1.000	1.000	1.000	1.000
15	1.000	1.049	1.000	1.049	1.000	1.000

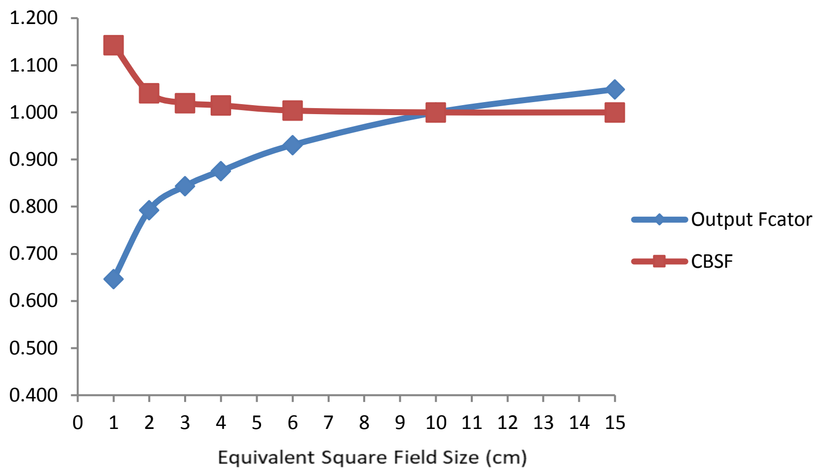

Figure 3. The output factor (OF) and collimator backscatter factor (CBSF) against the equivalent square field sizes

Table 3. The monitor units (MUs) and dose variations with the measured OF and corrected OF to deliver 12 Gy in a single fraction using the box field technique for various field sizes

Field size (cm^2)	MU		% Variation in MU	Dose measured (Gy)		% Variation in dose
	With measured OF	With corrected OF		With measured OF	With corrected OF	With correction
1×1	2879.3	2825.4	-1.9	12.25	12.05	0.4
2×2	2242.2	2219.0	-1.0	12.10	12.02	0.2
3×3	2068.7	2065.0	-0.2	12.01	12.00	0.0
4×4	1989.3	1983.5	-0.3	12.00	11.98	-0.2
8×8	1784.3	1779.9	-0.2	12.00	11.99	-0.1
10×10	1721.7	1719.3	-0.1	12.00	11.99	-0.1
15×15	1647.5	1644.6	-0.2	12.00	11.99	-0.1

Table 4. Variations in the CC01 IC-measured and AAA model-calculated point doses with and without the corrected output factors (OFs) for ten different targets by the DCA technique

Sr. No.	Target		D/fr (Gy)	Point dose variation (%)	
	Volume (cc)	Equivalent sphere diameter (mm)		AAAmeasured OF	AAAcorrected OF
P1	0.23	8	6.5	3.3	2.1
P2	0.36	9	16	3.1	1.9
P3	0.44	9	20	2.5	1.4
P4	0.49	10	20	2.3	1.2
P5	0.51	10	12	2.2	1.1
P6	0.54	10	13	1.9	1.3
P7	0.63	11	16	1.7	1.2
P8	0.81	12	20	1.5	1.0
P9	0.86	12	12	1.3	0.8
P10	2.1	16	6.5	1.1	0.6

The MU and dose variations with the measured OF and corrected OF to deliver 12 Gy in a single fraction, using the box field technique for various field sizes on the RW3 slab phantom, are shown in Table 3. There were 1.9%, 1%, and 0.2% reductions in MU with 1.8%, 0.9%, and 0.1% increases in the OF for field sizes of 1×1 cm^2, 2×2 cm^2, and 3×3 cm^2 defined for the jaws. The percentage deviation in dose was reduced with the corrected OF. However, we observed 0.1% to 0.3% reductions in the MU and 0.1% to 0.2% reductions in dose for uncorrected OFs in larger field sizes.

Deviations in the CC01-measured and AAA-calculated point doses, with and without the corrected OFs for ten different targets by the DCA technique, are shown in Table 4. The OF-corrected AAA dose algorithm showed a better agreement between the measured and simulated results. There was a 1.2% less deviation for a small target of 0.23 cc (8 mm equivalent sphere diameter) and a 0.5% less deviation for a target of 2.1 cc (16 mm equivalent sphere diameter) in a corrected OF beam model. No volume averaging correction was done in measurements with the CC01 CI.

Discussion

There is a general idea that TRS-483 CoP-recommended corrections mainly pertain to small fields used in CyberKnife® and TomoTherapy®, where the standard reference field does not exist. Huq et al. [22] carried out a dosimetric evaluation of the TRS483 CoP protocol in the Varian TrueBeam™ LINAC and compared it with the existing protocols. For the field OF determination, an unshielded diode detector (PTW 60017) and an Edge diode detector (Sun Nuclear) were used. The spread of measured data can be reduced for very small fields by multiplying the output correction factor by the ratio of readings. There was a 2% reduction in the mean value of the field OF in comparison with the uncorrected ratio of readings for an equivalent square field size of 1 cm.

For the 1-cm equivalent square field size, the correction factor decreased the OF from 0.734 to 0.726 for the PTW 60017 detector and from 0.752 to 0.729 for the Edge diode detector. In the present study, the OF was determined using the IBA CC01 CI, and correction increased the OF from 0.635 to 0.646. The TRS483 correction improved the field OF toward the MC-simulated value of 0.684. However, there are no studies on the clinical effects of TRS-483 CoP for small fields used in conventional LINACs.

We corrected the field OFs according to the TRS-483 recommendation. Changes in the OF as a function of field size were caused by changes in the phantom scatter, head scatter, and collimator backscatter into the monitor chamber. The photon beam source model and AAA volumetric dose calculation algorithm accounted for the phantom and head scatter effects. The remaining changes in the OF were assumed to be caused by the collimator backscatter (Eq. 7). After applying the correction, the OF increased, resulting in a decrease in the CBSF. This increased the gain of the dose monitor and altered the LINAC dose calibration of 1 cGy/MU. The AAA model considered these affects and reduced the MU. There was a 0.2% reduction in the MU for larger field sizes, without any correction. Therefore, apart from the CBSF calculation, the algorithm used small field OFs for some residual corrections in the beam modeling.

The AAA model-based algorithm has a few limitations and some approximations. It calculates the backscatter factor from the OFs, along with residual corrections to account for all phantom and head scatters [28, 29]. Therefore, the values are not expected to agree with the measured collimator backscatter factor. In the AAA algorithm, the configuration program optimizes the source model parameters using a dose deposition engine. The source model parameters are optimized using symmetric jaw delimited fields, but in TRS-483, corrections are provided for a dosimetric field size, which is defined by MLC. For small field treatments, Varian recommends to keep the jaw delimited field size at 3×3 cm^2 and to use MLC to shape the smallest target. For Varian, the LINAC MLC is located below the jaws; therefore, OFs of 1×1 cm^2 and 2×2 cm^2 (if not included) will not extremely affect the calculations if the jaws are positioned at 3×3 cm^2.

By keeping the jaws at 3×3 cm^2, we can prevent small field areas. This will compromise the dosimetric accuracy, and radiation leak through MLC in the absence of jaws will contribute to an out-of-field dose. If an OF up to 1×1 cm^2 is used in the source model parameter, there is an improvement in the dosimetric accuracy. To produce realistic CBSF, we need to use the MLC-delimited field output as input in the configuration program. However, this is not practical with AAA and Accuros XB algorithms. In this study, for OF measurements, readings were normalized to those in a 10×10 cm^2 field size. A field size of 3×3 cm^2 can be used for normalization, since it is sufficient to provide charge particle equilibrium [30]. However, a field size of 3×3 cm^2 is not always free from small field conditions. The choice of the detector is also important, as for the CC01 IC, this field size is not suitable.

Conclusion

This study provided further insights into the impact of TRS-483 CoP for small fields used in LINAC. There was no significant improvement (0.1%) in the beam quality correction factor for the reference dosimetry. The field OFs for small beams could be improved by incorporating the correction factors provided by the TRS-483. However, the extent of expected improvement depends on the source model of calculating algorithm and how well the generalized recommended corrections suit the user beam and the detector. The MC-simulated OF was a strong indicator for correcting the OFs for small fields. Nevertheless, the recommended correction was not sufficient for 1×1 cm^2, and the model did not include an OF for field sizes less than 1×1 cm^2. The TRS-483 corrections were provided for dosimetric field sizes defined by MLC, whereas the AAA algorithm source model parameters were optimized using jaw delimited fields. According to the Varian user guide, the AAA-calculated CBSF is not expected to agree with the measurements due to model limitations. Although there was a better agreement between the measured and simulated doses for small fields with the corrected OF table, deviations in point dose for the patient-specific QA of DCA plans reduced for the SRS/SRT cases.

Acknowledgment

The authors sincerely thank Ms. Sandhya Madhvapathi Rao for her support throughout the manuscript preparation process.

References

1. Georg D, Knöös T, McClean B. Current status and future perspective of flattening filter free photon beams. Med Phys. 2011 Feb 15; 38:1280‑93.
2. Lang S, Shrestha B, Graydon S, Cavelaars F, Linsenmeier C, Hrbacek J;, et al. Clinical application of flattening filter free beams for extracranial stereotactic radiotherapy. Radiother Oncol. 2013 Feb;106(2):255-259.
3. Mamballikalam G, Senthilkumar S, Jaon bos RC, Ahamed Basith PM, Jayadevan PM. Stereotactic radiotherapy for small and very small tumours (≤1 to ≤3 cc): Evaluation of the influence of volumetric-modulated arc therapy in comparison to dynamic conformal arc therapy and 3D conformal radiotherapy as a function of flattened and unflattened beam models. J Radiother Pract. 2020 Jan 13, 1-7.
4. Andreo P, Burns DT, Hohlfeld K, Huq MS, Kanai T,Laitano F,; et al. Absorbed dose determination in external beam radiotherapy: An international code of practice for dosimetry based on standards of absorbed dose to water. Vienna: IAEA TRS-398; 2001.
5. Almond PR, Biggs PJ, Coursey BM, Hanson WF, Huq MS, Nath R;, et al. AAPM’s TG-51 protocol for clinical reference dosimetry of high-energy photon and electron beams. Med. Phys.1999 Sep 1; 26:1847‑1870.
6. Derreumaux S, Etard C, Huet C, Trompier F, Clairand I, Bottollier JF;, et al. Lessons from recent accidents in radiation therapy in France. Radiat. Prot. Dosim. 2008 Aug 25; 131:130‑135.
7. Das IJ, Ding GX, Ahnesjö A. Small fields: nonequilibrium radiation dosimetry. Med Phys. 2008 Dec 20;35(1):206-215.
8. Mamballikalam G, Senthilkumar S, Jayadevan PM, Jaon bos RC, Ahamed Basith PM, Inippully R,; et al. Evaluation of dosimetric parameters of small fields of 6 MV flattening filter free photon beam measured using various detectors against Monte Carlo simulation. J Radiother Pract. 2020 March 9; 1-8.
9. Francescon P, Cora S, Cavedon C. Total scatter factors of small beams: a multidetector and Monte Carlo study. Med Phys. 2008 Jan 15; 35:504‑13.
10. Alfonso R, Andreo P, Capote R, Huq MS, Kilby W, Kjäll P;, et al. A new formalism for reference dosimetry of small and nonstandard fields. Med Phys. 2008 Nov;35(11):5179-5186.
11. Pantelis E, Moutsatsos A, Zourari K, Kilby W, Antypas C, Papagiannis P,; et al. On the implementation of a recently proposed dosimetric formalism to a robotic radiosurgery system. Med Phys. 2010 Apr 29;37(5):2369‑79.
12. Ralston A, Liu P, Warrener K, McKenzie D, Suchowerska N. Small field diode correction factors derived using an air core fibre optic scintillation dosimeter and EBT2 film. Phys Med Biol. 2012 Apr 13; 57:2587‑602.
13. Lechner W, Palmans H, Sölkner L, Grochowska P, Georg D. Detector comparison for small field output factor measurements in flattening filter free photon beams. Radiother Oncol. 2013 Nov 18;109(3):356-360.
14. Benmakhlouf H, Sempau J, Andreo P. Output correction factors for nine small field detectors in 6 MV radiation therapy photon beams: a PENELOPE Monte Carlo study. Med Phys. 2014 Apr 1; 41(4):041711.
15. Azangwe G, Grochowska P, Georg D, Izewska J , Hopfgartner J, Lechner W,; et al. Detector to detector corrections: a comprehensive experimental study of detector specific correction factors for beam output measurements for small radiotherapy beams. Med Phys. 2014 Jun 20;41(7):072103.
16. Aspradakis MM, Byrne JP, Palmans H, Duane S, Conway J, Warrington AP,; et al. Small field MV photon dosimetry. IPEM report 103. 2010; IAEA: CN-182.
17. Das IJ, Morales J, Francescon P. Small field dosimetry: What have we learnt? AIP Conference Proceedings; 2016; 1747:060001.
18. Scott AJD, Nahum AE, Fenwick JD. Monte Carlo modeling of small photon fields: quantifying the impact of focal spot size on source occlusion and output factors, and exploring miniphantom design for small-field measurements. Med Phys. 2009 Jun 12; 36: 3132‑44.
19. Scott AJD, Kumar S, Nahum AE, Fenwick JD. Characterizing the influence of detector density on dosimeter response in non-equilibrium small photon fields. Phys Med Biol. 2012 Jun 22; 57: 4461‑76.
20. Fenwick JD, Kumar S, Scott AJD, Nahum AE. Using cavity theory to describe the dependence on detector density of dosimeter response in non-equilibrium small fields. Phys Med Biol. 2013 Apr 11;58: 2901‑23.
21. IAEA, AAPM. Dosimetry of small static fields used in external beam radiotherapy: an international code of practice for reference and relative dose determination. Technical reports series no. 483. Vienna: IAEA; 2017.
22. Huq MS, Hwang MS, Teo TP, Jang SY, Heron DE, Lalonde RJ. A dosimetric evaluation of the IAEA-AAPM TRS 483 code of practice for dosimetry of small static fields used in conventional linac beams and comparison with IAEA TRS-398, AAPM TG 51, and TG 51 Addendum protocols. Medical physics. 2018 Sep;45(9):4257-73.Huq MS, Hwang MS, Teo TP, Jang SY, Heron DE, Lalonde RJ. A dosimetric evaluation of the IAEA-AAPM TRS483 code of practice for dosimetry of small static fields used in conventional linac beams and comparison

with IAEA TRS-398, AAPM TG51, and TG51 Addendum protocols. Med Phys. 2018 Jul 15;10.1002/mp.13092.
23. British Institute of radiology. Central axis depth dose data for use in radiotherapy. BJR Supplement 25. London: British institute of radiology; 1996.
24. Khan FM, Gibbons JP. The physics of radiation therapy. 5th ed. Philadelphia: Lippincott Williams & Wilkins; 2014.
25. Rodriguez M, Sempau J, Brualla L. PRIMO: a graphical environment for the Monte Carlo simulation of Varian and Elekta linacs. Strahlenther Onkol. 2013 Sep 6;189(10):881-886.
26. Gete E, Duzenli C, Milette MP, Mestrovic A, Hyde D, Bergman AM;, et al. A Monte Carlo approach to validation of FFF VMAT treatment plans for the TrueBeam linac. Med Phys. 2013 Jan 15;40(2):021707.
27. Belosi MF, Rodriguez M, Fogliata A, Cozzi L Sempau J , Clivio A,; et al. Monte Carlo simulation of TrueBeam flattening-filter-free beams using varian phase-space files: comparison with experimental data. Med Phys. 2014 Apr 17;41(5):051707.
28. Sievinen J, Ulmer W, Kaissl W. AAA photon dose calculation model in Eclipse. Palo Alto (CA): Varian Medical Systems; 2005. [RAD #7170B].. 2005;118:2894.
29. Torsti T, Korhonen L, Petäjä V. Using Varian photon beam source model for dose calculation of small fields. Clinical Perspectives. Helsinki, Finland: Varian Medical Systems; 2013.
30. Li XA, Soubra M, Szanto J, Gerig LH. Lateral electron equilibrium and electron contamination in measurements of head-scatter factors using miniphantoms and brass caps. Med Phys. 1995 Jul;22(7):1167-1170.