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Investigation of field output factors using IAEA-AAPM TRS-483 code of practice recommendations and Monte Carlo simulation for 6 MV photon beams

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Original Article

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Abstract

Introduction: This study aims to experimentally determine field output factors using the methodologies suggested by the IAEA-AAPM TRS-483 for small field dosimetry and compare with the calculation from Monte Carlo (MC) simulation.
Methods: The IBA-CC01, Sun Nuclear EDGE and IBA-SFD detectors were employed to determine the uncorrected and the corrected field output factors for 6 MV photon beams. Measurements were performed at 100 cm source to axis distance, 10 cm depth in water, and the field sizes ranged from 1×1 to 10×10 cm^2. The use of field output correction factors proposed by the TRS-483 was utilised to determine field output factors. The measured field output factors were compared to that calculated using the egs_chamber user code.
Results: The decrease in the percentage standard deviation of the measured three detectors was observed after applying the field output correction factors. Measured field output factors using CC01 and EDGE detectors agreed with MC values within 3% for field sizes down to 1×1 cm^2, except the SFD detector.
Conclusions: The corrected field output factors agree with the calculation from MC, except the SFD detector. CC01 and EDGE are suitable for determining field output factors, while the SFD may need more implementation of the intermediate field method.

Introduction

Accurate determination of field output factors ($\Omega^{f_{\text{clin}},f_{\text{msr}}}_{Q_{\text{clin}},Q_{\text{msr}}}$) in external beam radiotherapy is critical to transfer the absorbed dose in water from reference field size to other clinical field sizes. It is necessary for the commissioning of treatment planning systems that use these fields. The field output factors are defined as the ratio of absorbed dose in water in any clinical field size $\left(D^{f_{\text{clin}}}_{w,Q_{\text{clin}}}\right)$ to absorbed dose in water for machine-specific reference field size $\left(D^{f_{\text{msr}}}_{w,Q_{\text{msr}}}\right)$, used for clinical reference dosimetry.

For large fields, the field output factors can be estimated by the ratio of the detector reading in the clinical field ($M^{f_{\text{clin}}}_{Q_{\text{clin}}}$) to that of machine-specific reference field ($M^{f_{\text{msr}}}_{Q_{\text{msr}}}$). However, the $M^{f_{\text{clin}}}_{Q_{\text{clin}}}/M^{f_{\text{msr}}}_{Q_{\text{msr}}}$ is not an accurate determination of field output factors in a small field because of many factors. The most significant factors are small field detector perturbations and volume averaging over the detector's sensitive volume and the difference in density between detector sensitive material and water. In addition, the effective atomic number of surrounding detector construction materials can affect the perturbation factors.[1–4]

Numerous studies have shown a significant variation in the ratio of reading when using different types of small field detectors, especially in very small field sizes.[5–8] Moreover, the discrepancies in the ratio of reading increased when the field sizes turned smaller. The previous lesson of an accidental overdosage for beams defined by the Brainlab m3 micro-multileaf collimator (MLC) was published due to the use of unsuitable detectors for measuring field output factors without additional corrections.[9] To address this problem, the ratio of reading of appropriate detector should be corrected by the field output correction factors $\left(k^{f_{\text{clin}},f_{\text{msr}}}_{Q_{\text{clin}},Q_{\text{msr}}}\right)$ introduced in the formalism by Alfonso et al.[10]

$$\Omega_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}} = \frac{M_{Q_{clin}}^{f_{clin}}}{M_{Q_{msr}}^{f_{msr}}} \, k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}} \quad (1)$$

Field output correction factors ($k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$) for many active detectors were published in the IAEA-AAPM TRS-483 code of practice. This publication is the guideline for dosimetry procedures of small static fields in external beam radiotherapy.[1]

A dosimetric evaluation of the IAEA-AAPM TRS-483 code of practice for small static fields was conducted by Huq et al.[11] The same year, an evaluation of the IAEA-AAPM TRS-483 protocol of dosimetry in stereotactic cones using several detectors was conducted.[12] Both studies suggested that the application of the protocol improved consistency in determining small field output factors. A recent clinical implementation of IAEA-AAPM TRS-483 was performed by Mamesa et al.[8] Their study was performed for commissioning stereotactic radiosurgery. They reported that the corrected field output factors based on IAEA-AAPM TRS-483 reduced the discrepancies of monitor units (MUs) among the difference detector commissioning data for the treatment planning system.

Aside from direct measurement, Monte Carlo (MC) simulation has been regarded as an alternative tool for determining field output factors[5,13,14] and field output correction factors[2,15–19] in small field dosimetry. The performance of MC calculated output factors hinges on the commissioning of a beam model based on measurements. This study aims to compare measured field output factor data directly to MC simulations of field output factors using a tuned MC beam model.

Materials and Methods

Determination of measured field output factors

The measurements were performed on the Varian TrueBeam linear accelerator (Varian Medical System, Palo Alto, CA, USA) with 6 MV WFF (with flattening filter). The $TPR_{20,10}$ as the 6 MV beam quality of this machine was 0·667. All measurements were conducted in Blue Phantom (IBA Dosimetry, Nuremberg, Germany) at 10 cm depth of measurement and 100 cm source to axis distance (SAD). The geometric field sizes ranging from 1×1 to 10×10 cm^2 were defined by jaws because the output factors used for commissioning in Eclipse treatment planning (Varian Medical System, Palo Alto, CA, USA) were measured from this configuration. Therefore, it is more convenient to use jaw field sizes in this study. Moreover, there was no difference in output factors between field sizes defined by the jaws and MLC (field sizes were defined by a square MLC aperture while the jaws were slightly retracted with 0·5 cm open larger than the leaves).[8] The machine-specific reference field size was 10×10 cm^2.

Three small active detectors examined in this work were CC01 ionisation chamber (IBA Dosimetry, Nuremberg, Germany), EDGE detector (Sun Nuclear, Melbourne, FL, USA) and SFD unshielded diode (IBA Dosimetry, Nuremberg, Germany). Table 1 shows the dimension and physical properties of each detector. All detectors were attached to the holder of a water phantom scanning system. Detector orientation with respect to the central axis of the beam was set following the guidelines in the IAEA-AAPM TRS-483 code of practice. The sensitive volume of CC01 and EDGE were positioned in a perpendicular direction towards the beam direction, while SFD was orientated in the parallel direction, as shown in Figure 1.

The beam profile scanning was performed to determine the position of the sensitive volume of the detector at the centre of the beam axis. The scanning for each detector was conducted in 2×2 cm^2 field size. Before measurement, the centre area of the active detector was aligned with the beam centre. Then, each field size was repeatedly measured three times.

The equivalent square field sizes (S_{clin}) were also determined in this study. They were utilised for selecting of the $k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$. Equivalent square field size is given by $\sqrt{X.Y}$, where X and Y are cross-plane and in-plane full width at half maximum (FWHM), respectively, under measured conditions of 100 cm SAD and 10 cm depth. The FWHM in both cross-plane and in-plane were acquired by scanning the beam profile using the EDGE diode detector, a small detector suitable for beam scanning.

The ratio of the readings (uncorrected field output factors) of CC01, EDGE and SFD in the small field relative to the reference field was determined by $M_{Q_{clin}}^{f_{clin}}/M_{Q_{msr}}^{f_{msr}}$ for all field sizes.

The ratio of readings was directly corrected by $k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ based on Equation 1. The $k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ as a function of S_{clin} was obtained from Table 26 in IAEA-AAPM TRS-483 code of practice.[1] The dosimeter reading uncertainty at machine-specific reference field and clinical fields, the uncertainty of $k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ [1] and the effect of the set-up and jaw positioning uncertainties of 1·3%[20] were quadratically summed to determine the total uncertainty of the field output factors.

Determination of calculated field output factors (MC simulation)

The EGSnrc code[21] was used in this study. BEAMnrc was employed for modelling the TrueBeam linear accelerator with 6 MV flattened photon beams. The treatment head geometry of TrueBeam is not available. The previous study claimed a considerable similarity of head assembly geometries between the Clinac 2100 CD and the TrueBeam linacs.[22] Therefore, this study utilised the treatment head geometry from Clinac 2100 CD, geometry data provided by the manufacturer. Figure 2 shows the schematic of the accelerator model that consists of a target, primary collimator, vacuum window, flattening filter, monitor chamber, mirror and jaws. The beam model used in this study has been previously validated for 6 MV flattened beams by comparing the calculated beam profile and depth dose (DD) with the measurements.[23] The source parameters used for modelling were 5·9 MeV initial electron energy and 0·11 cm FWHM. For a field size ranging from $0{\cdot}5 \times 0{\cdot}5$ to 10×10 cm^2, the average difference between measured and calculated DD was 0·94%. The average difference between measured and calculated beam profiles at 10 cm depth was 0·57%. For beam source setting in egs_chamber user code, the BEAM accelerator code was compiled as a shared library and provided with its input file and pegs file.

The absorbed dose in the small volume of water was determined by the egs_chamber user code. A water phantom with a dimension of $30 \times 30 \times 30$ cm^3 was created. The dose scoring volume was generated in a cylindrical shape. The scoring volume dimension was diversified following the field size to reduce the simulation time in larger field sizes. The dimension of 0·015, 0·03 and 0·1 cm radius was set for field size up to 2×2, 3×3 to 4×4 and 6×6 to 10×10 cm^2, respectively. The length of the scoring volume was set to 0·05 cm for all field sizes. The cylindrical volume was placed parallel to the central beam axis at 100 cm SAD (10 cm depth and 90 cm source to surface distance).

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Table 1. Lists of detectors were used in this study.

Small field detectors	Materials	Z_{eff} of sensitive volume	Density (g/cm^3)	Diameter/side length (mm)	Thickness (mm)	Sensitive volume (mm^3)
IBA/Wellhöfer CC01	Air Wall: C-552 Central electrode: steel	7·6	0·0012[a]	2	3·6	10 (0·01 cm^3)
Sun Nuclear EDGE detector	Silicon	14	2·33	0·8	0·03	0·019
IBA SFD unshielded diode	Silicon	14	2·33	0·6	0·06	0·017

[a]Dry air at 20 °C and 101.3 kPa.

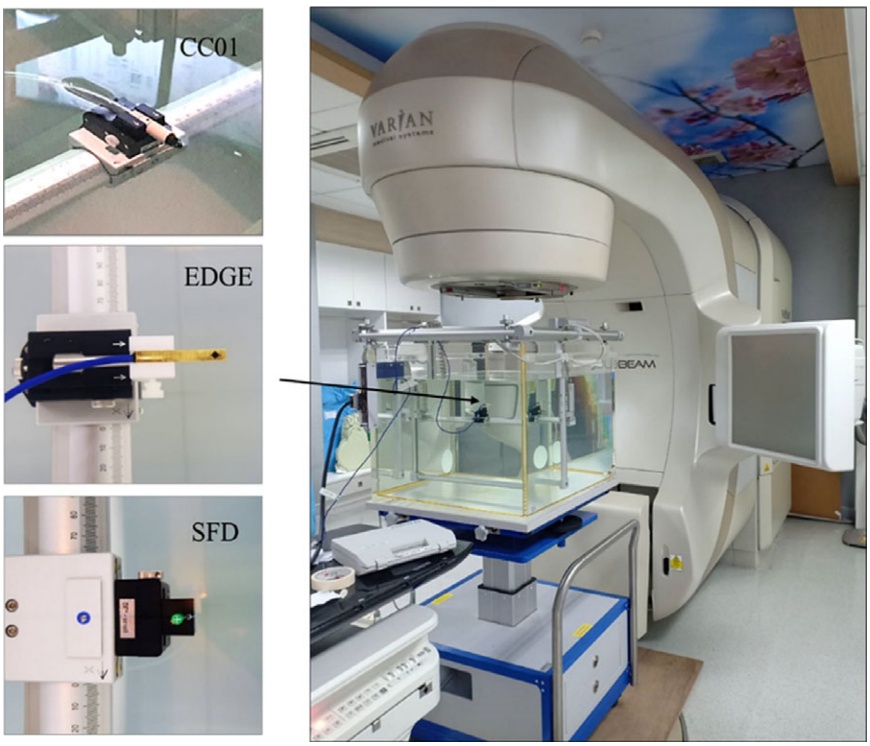

Figure 1. The measurement set-up of the small detector in blue water phantom with perpendicular orientation for CC01 and EDGE and parallel orientation for SFD.

The parameters for particle transport are ECUT = 512 keV and PCUT = 10 keV. Variance reduction techniques such as Russian roulette range rejection and photon cross-section enhancement were implemented. The number of particles in a scoring volume of water was calculated to reach the statistical uncertainty of 0·05–0·08% based on field sizes. The doses in small water volume (cGy/particle) were determined for field sizes ranging from 1×1 to 10×10 cm^2. Later, the field output factors were calculated with the 10×10 cm^2 as the reference field.

Data analysis

The percentage standard deviation (%SD) for each field size was determined to evaluate the variation of output factors among different detectors. It is calculated as the standard deviation (SD) ratio to the mean, expressed as a percentage. The percentage differences between measured and calculated field output factors were investigated for an individual detector to assess improvement in consistency of field output factors after implementing the $k^{f_{clin},f_{msr}}_{Q_{clin},Q_{msr}}$.

Results

The measured field output factors

The ratio of readings among the three detectors is tabulated in Table 2. For intermediate field size (4×4 to 8×8 cm^2), the ratio of readings between CC01 and EDGE was comparable. On the other side, SFD exhibited the lowest value and deviated from other detectors. For 3×3 cm^2 field size or less, the highest ratio of reading was shown by EDGE shielded diode detector. The lowest ratio of reading was observed in the SFD. Comparing the data among the three detectors, the %SD increased when field sizes became smaller. The highest %SD of 4·3% was obtained in 1×1 cm^2 field size.

As presented in Table 2, the corrected field output factors $\left(\Omega^{f_{clin},f_{msr}}_{Q_{clin},Q_{msr}}\right)$ showed much better consistency among three detectors with %SD of 1·1% in the 1×1 cm^2 field size ($S_{clin} = 9{\cdot}6$ mm). After correction, the $\Omega^{f_{clin},f_{msr}}_{Q_{clin},Q_{msr}}$ of EDGE agreed well with CC01 in small fields to within 1·3%. However, the $\Omega^{f_{clin},f_{msr}}_{Q_{clin},Q_{msr}}$ of SFD was still lower compared to the other detectors.

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Table 2. The ratios of reading, and measured and calculated field output factors as a function of equivalent square field size, S_{clin}.

Side of square field (cm)	10	8	6	4	3	2	1
S_{clin} (cm)	9·98	7·98	5·97	3·96	2·96	1·96	0·96
Detector	Ratio of reading						
CC01	1·000	0·963	0·917	0·859	0·825	0·784	0·665
EDGE	1·000	0·964	0·919	0·863	0·831	0·796	0·713
SFD	1·000	0·953	0·900	0·836	0·801	0·759	0·662
Average	1·000	0·960	0·912	0·853	0·819	0·780	0·680
SD	0·000	0·006	0·010	0·015	0·016	0·019	0·029
%SD	0·00	0·60	1·14	1·72	1·96	2·41	4·27
	Field output factors						
CC01	1·000	0·965	0·920	0·865	0·831	0·791	0·678
EDGE	1·000	0·964	0·919	0·863	0·830	0·791	0·687
SFD	1·000	0·961	0·915	0·857	0·824	0·783	0·672
Average	1·000	0·963	0·918	0·862	0·829	0·789	0·679
SD	0·000	0·002	0·003	0·004	0·004	0·004	0·008
%SD	0·00	0·20	0·28	0·50	0·48	0·56	1·11
	Calculated field output factors						
MC	1·000	0·968	0·927	0·875	0·843	0·801	0·699

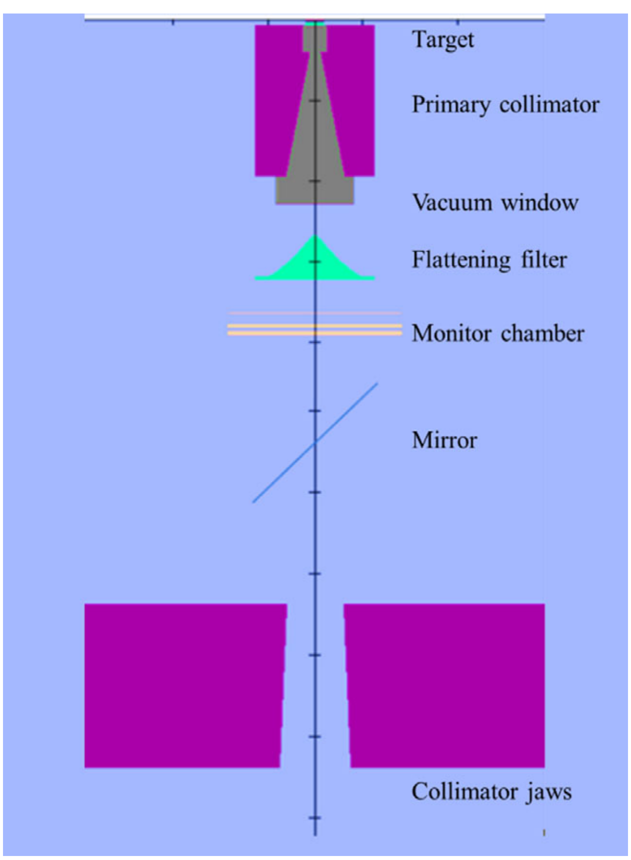

Figure 2. Schematics of treatment head geometry for a linear accelerator.

The dosimeter reading uncertainties at a machine-specific reference field and clinical fields were within 0·2% for all detectors and all field sizes. The uncertainties of $k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ of CC01, EDGE and SFD were 1·1, 0·7 and 0·5%, respectively. After combining all uncertainties with the effect of the set-up and jaw positioning uncertainties of 1·3%[20], the total uncertainties of field output factors determination with expansion were within 3% (k = 2) for all detectors and all field sizes, except the CC01 in 1×1 cm^2 field size.

Comparison between measured and calculated field output factors

The calculated field output factors from the MC simulation are shown in Table 2, and they were compared with that of measurement. The percentage difference between measurement and calculation by MC simulation in particular methods for three detectors is reported in Figure 3. A discrepancy is expected for uncorrected output factors because the MC calculations report dose to water ratios. In contrast, the measured detector reading ratios include detector response perturbations in the small field.

A decrease in percentage difference to a calculation by MC simulation was found after applying the $k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ for all detectors, as shown in Figure 3. For field size larger than 1×1 cm^2, the percentage differences were within 1·5% for EDGE and CC01 detectors and 2·5% for SFD. For 1×1 cm^2, a difference of 1·7 and 3·0% was observed in EDGE and CC01 detectors, respectively. Meanwhile, a percentage difference of 3·8% was detected in the SFD detector.

Discussion

For the small field size, the highest ratio of reading was exhibited by EDGE shielded diode detector due to the high density of sensitive volume material in conjunction with the embedded brass material as its shielding. The presence of brass shielding material increases the fluence of secondary electrons in silicon diode due to the higher mass-energy absorption coefficient of brass. Thus, an over-response was observed for the EDGE detector in the small field, which agrees with previous publications.[24–26]

For intermediate and large field sizes, the ratio of readings of CC01 and EDGE was comparable, while the results of SFD were lower and showed deviation from other detectors. The unshielded diode exhibited an over-response of output in large fields due to low energy scattered photons in the beam. Hence, an underestimation result was observed when the readings were normalised to a large field of 10×10 cm^2. This over-response, however, did not occur in the EDGE shielded diode detector since shielding material is present that minimises the effect on the detector response from low energy scattered photons. This explains the agreement in the ratio of readings between CC01 and EDGE for intermediate and large field sizes.

The uncertainty of CC01 was higher than 3% (k = 2) in 1×1 cm^2. The significant uncertainty was affected by the field output correction factor of CC01 (1·1%). However, these outcomes agreed with the results presented in Tolabin et al. study.[27]

In Table 2, there was a significant variation of the ratio of readings among different types of detectors for smaller field sizes. Therefore, the implementation of field output correction factors is recommended. Preferably, the IAEA-AAPM TRS-483 code of practice stated that averaging the field output factors from at least two different types of suitable small detectors is advised according to the French Society of Medical Physics (SFPM).[1]

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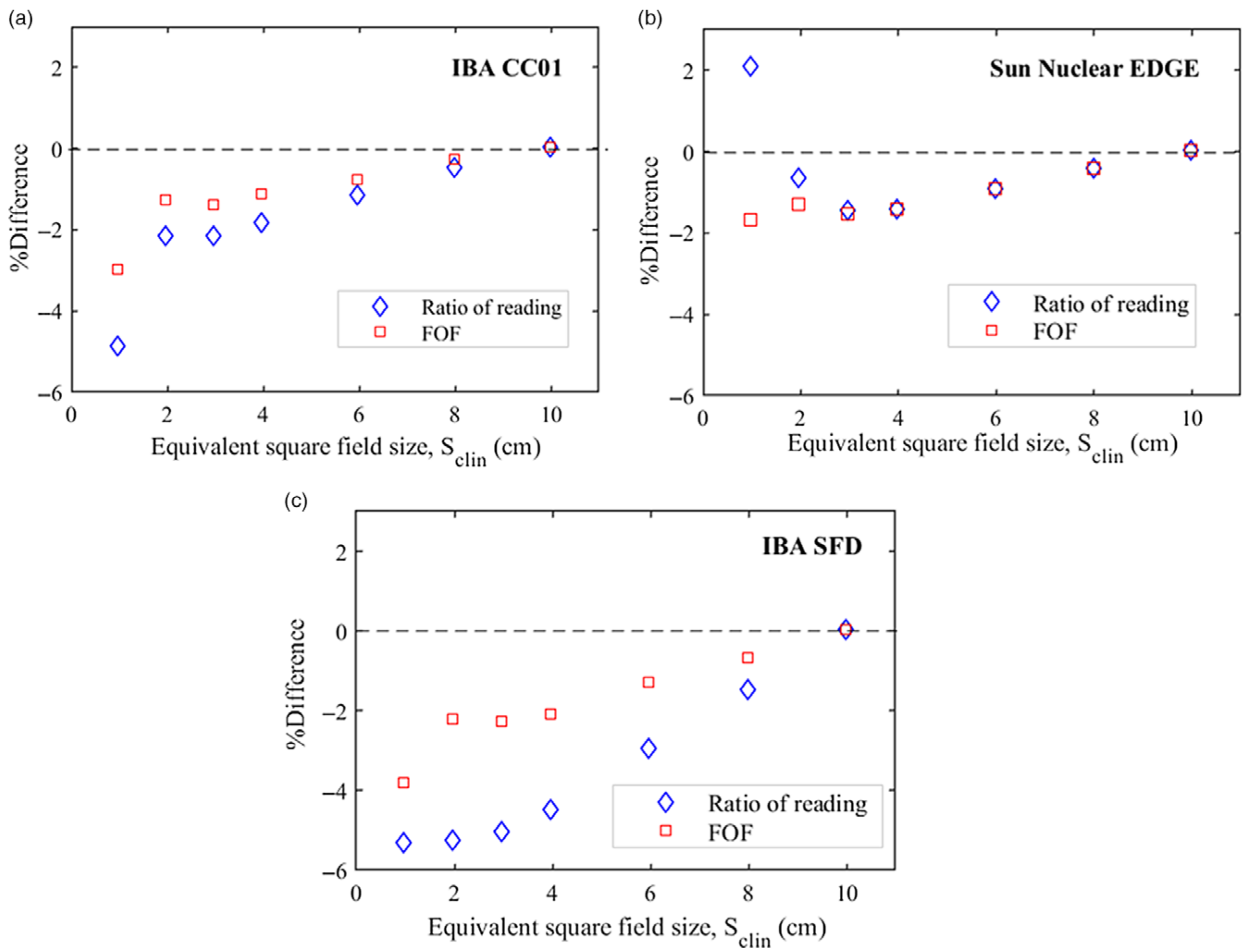

Figure 3. The percentage difference between the measured field output factors and calculated field output factors for CC01, EDGE and SFD detectors (FOF stands for field output factors).

The larger difference between MC calculated field output factors and measured values following the IAEA-AAPM TRS-483 code of practice recommendations for the smallest field size (10 mm) might be due to a combination of reasons. These include sensitivity of the tuned MC model to exact source and collimator parameters, which affects the prediction of accelerator output in very small fields (10 mm). In addition, experimentally, detector positioning during the measurements in the smallest field size and alignment of the detector sensitive volume with the actual beam axis is a challenge. For this reason, the IAEA-AAPM TRS-483 report recommends the use of multiple detectors corrected using field output correction factors to establish the field output factor for small fields.

In practice, accurate field output factors are needed to implement in the computerised treatment planning system. When a small radiation field is used, the accurate determination of field output factors is challenging. This study provided the data to consider some suitable detectors for small field output factor determination. The result also confirms that the small detectors with appropriate field output correction factors are necessary to achieve the accurate field output factors, as shown in the results compared with MC simulation. The study of Mamesa et al. supports our results. They illustrated that the accuracy of the field output factors is one of the parameters influencing the dose calculation in the treatment planning system in terms of MU calculation for clinical cases of Intensity modulated radiotherapy and Volumetric modulated radiotherapy.[8] Therefore, selecting the appropriate detectors with the recommended field output correction factors is essential for beam commissioning.

In addition, the CC01 and EDGE diode detectors with applying the field output correction factors are recommended for determining the field output factors for field size down to 1×1 cm^2. The SFD unshielded diode detector showed higher field size dependence and over-response of output in large field sizes. Therefore, the intermediate field method to link the difference of response for small and large fields for unshielded diode detectors should be applied for SFD as suggested in IAEA-AAPM TRS-483.[1]

Conclusion

The determination of field output factors in small fields using the field output correction factors based on the IAEA-AAPM TRS-483 code of practice presents a lower %SD (better consistency of three different detectors). The measured field output factors agree well with the calculated field output factors using a tuned MC beam model (less than 3% difference), except the smallest field size of SFD unshielded diode detector. For further recommendation,

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CC01 and EDGE diode detector are suitable for determining field output factors, while the SFD may need more implementation of the intermediate field method as the suggestion in IAEA-AAPM TRS-483.

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Conflict of Interest. No conflict of interest to declare.

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