

Technical Report: Evaluation of peripheral dose for flattening filter free photon beams

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Purpose: To develop a comprehensive peripheral dose (PD) dataset for the two unflattened beams of nominal energy 6 and 10 MV for use in clinical care.

Methods: Measurements were made in a $40 \times 120 \times 20$ cm³ (width \times length \times depth) stack of solid water using an ionization chamber at varying depths (d_{\max} , 5, and 10 cm), field sizes (3×3 to 30×30 cm²), and distances from the field edge (5–40 cm). The effects of the multileaf collimator (MLC) and collimator rotation were also evaluated for a 10×10 cm² field. Using the same phantom geometry, the accuracy of the analytic anisotropic algorithm (AAA) and Acuros dose calculation algorithm was assessed and compared to the measured values.

Results: The PDs for both the 6 flattening filter free (FFF) and 10 FFF photon beams were found to decrease with increasing distance from the radiation field edge and the decreasing field size. The measured PD was observed to be higher for the 6 FFF than for the 10 FFF for all field sizes and depths. The impact of collimator rotation was not found to be clinically significant when used in conjunction with MLCs. AAA and Acuros algorithms both underestimated the PD with average errors of -13.6% and -7.8% , respectively, for all field sizes and depths at distances of 5 and 10 cm from the field edge, but the average error was found to increase to nearly -69% at greater distances.

Conclusions: Given the known inaccuracies of peripheral dose calculations, this comprehensive dataset can be used to estimate the out-of-field dose to regions of interest such as organs at risk, electronic implantable devices, and a fetus. While the impact of collimator rotation was not found to significantly decrease PD when used in conjunction with MLCs, results are expected to be machine model and beam energy dependent. It is not recommended to use a treatment planning system to estimate PD due to the underestimation of the out-of-field dose and the inability to calculate dose at extended distances due to the limits of the dose calculation matrix. © 2016 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4958963>]

Key words: peripheral dose, out-of-field dose, flattening filter free, unflattened beams

1. INTRODUCTION

As the clinical use of the flattening filter free (FFF) photon beams increases in radiation therapy, the need for in-depth characterizations of FFF beams has arisen due to inherent differences in beam properties from the conventional flattened beams. The reported benefits of unflattened compared to flattened photon beams include an increase in dose rate, reduced scatter and leakage radiation, and a reduction in peripheral dose (PD).^{1–11} Peripheral or out-of-field dose is defined as the absorbed dose outside of the radiation treatment field borders.^{12,13} Peripheral dose is due to contributions from leakage radiation from the head of the treatment unit, scatter radiation from the collimator head and beam modifiers, and internal scatter within the patient.^{14–16} The PD may be clinically significant when dose to the fetus or to organs with low dose tolerances (e.g. gonads, lens of eye)

must be minimized. Peripheral doses are also relevant when considering the risk of secondary malignancies or the potential failure of the implanted electronic devices (e.g., pacemakers, defibrillators).^{5,12,15,17–19}

Numerous studies have investigated the trends in PD for flattened photon beams with varying field sizes, depth, and distances from the field edge, as well as the effects of beam modifiers [e.g., multileaf collimators (MLCs)] and collimator rotation.^{12,14–16,20–24} The removal of the flattening filter changes the profile and dosimetric characteristics of photon beams.³ As such, one cannot assume that the PD trends observed with flattened beams will be consistent with unflattened beams. Due to known inaccuracies of PD calculations with commercial treatment planning systems (TPSs),^{18,25} the typical means of assessing PD are through phantom measurements or Monte Carlo (MC) simulations.^{4,6}

Several MC publications^{2,5,7,9} have investigated the PD of unflattened photon beams compared to flattened beams, both in the scenario in which the incident electron energy of the flattened and unflattened beams is matched (i.e., similar to the model utilized by Varian Medical Systems¹⁰), and when the incident energy of the unflattened beam is increased to match the attenuation of the corresponding flattened beam (i.e., similar to the model employed by Elekta¹⁰). The majority of these publications^{2,7,9} have suggested that the PDs of unflattened beams are lower than the corresponding PDs for flattened beams, and that this difference increases with photon energy and decreasing field size. Additionally, Almqvist *et al.*⁷ simulated both 6 FFF scenarios (incident electron energies of 6.45 and 8 MeV), and reported that the PD reduction was enhanced by increasing the incident photon energy (i.e., 8 MeV), suggesting that PD is not only dependent on the machine design but also on the implementation of the FFF technology. In contrast to previously published MC simulations, Kry *et al.*⁵ reported that the difference in PD between flattened and unflattened photon beams is dependent on the distance from the field edge. Within the first 3 cm of the field, they observed that the PD of the FFF beam was lower than the corresponding flattened beam. At distances between 3 and 15 cm from the field edge, the PD was typically higher in FFF mode, and beyond 15 cm, the PD was again lower in FFF mode.

Also, PD of FFF beams has been studied for the treatment plans generated for the intensity modulated radiation therapy (IMRT), stereotactic body radiotherapy (SBRT), and stereotactic radiosurgery (SRS)^{4,8} as well as for square and rectangular open fields.^{7,11,19} Similar to the MC studies, these publications have suggested that the PD of unflattened beams is lower than flattened beams of the same nominal energy and that the relative difference in PD increases with energy.

The intention of the current study was to develop a comprehensive PD dataset for 6 FFF and 10 FFF photon beams that could be used for scenarios where the PD may be clinically relevant. We also compared the trends in PD measurements between the two unflattened beams, as well as compared the PD of the 6 FFF beam with the published results of a 6 MV flattened beam.²⁶

2. METHODS

All measurements were performed for the 6 and 10 MV unflattened photon beams of a Varian TrueBeam linac (Varian Medical Systems, Palo Alto, CA) running version 1.6 of the control software and equipped with the Millennium 120-leaf MLCs. The PD of the two unflattened beams was determined with a $40 \times 120 \times 20 \text{ cm}^3$ (width \times length \times depth) stack of solid water (Solid Water model 457, Gammex/RMI, Milwaukee, WI) with an Exradin A12 ionization chamber (Standard Imaging, Middleton, WI) inserted in a 2 cm slab of solid water when positioned on the treatment couch (see Fig. 1). Measurements were acquired at various depths [nominal depth of maximum dose, d_{max} (1.5 cm for the 6 FFF beam and 2.0 cm for the 10 FFF beam), 5, and 10 cm], field sizes (3×3 , 6×6 , 10×10 , 15×15 , 20×20 , 25×25 , and $30 \times 30 \text{ cm}^2$), and distances (5, 10, 15, 20, 30, and 40 cm) between the field edge and the center of the ionization chamber. The field aperture was shaped by matching the MLCs and the jaws, and a 90° collimator rotation was used. The distance from the field edge was defined at the surface of the solid water phantom where the source-to-surface distance was 100 cm (see Fig. 1). Peripheral dose measurements were normalized to the measured central axis (CAX) dose at the nominal d_{max} for a given field size. Additionally, the effects of the tertiary MLC and collimator rotation were evaluated for a $10 \times 10 \text{ cm}^2$ field size by acquiring PD measurements with the collimator positioned at both 0° and 90° for the field aperture set with the collimator jaw and/or MLCs.

To compare the measured PD with the dose calculated in the TPS, a phantom with the same dimensions as the measurement phantom was generated in ECLIPSE (version 10, Varian Medical Systems, Palo Alto, CA). Beam models for the 6 FFF and 10 FFF were developed using the Varian TrueBeam representative dataset which was based on the average of three machines.²⁷ A series of treatment plans were created using a single anterior beam corresponding to each of the measured, static field sizes with a 90° collimator rotation. To calculate the PDs, reference points were added at the same depths and distances from the field sizes that were measured. The PDs were calculated using both the analytic anisotropic algorithm (AAA) version 10.0.42

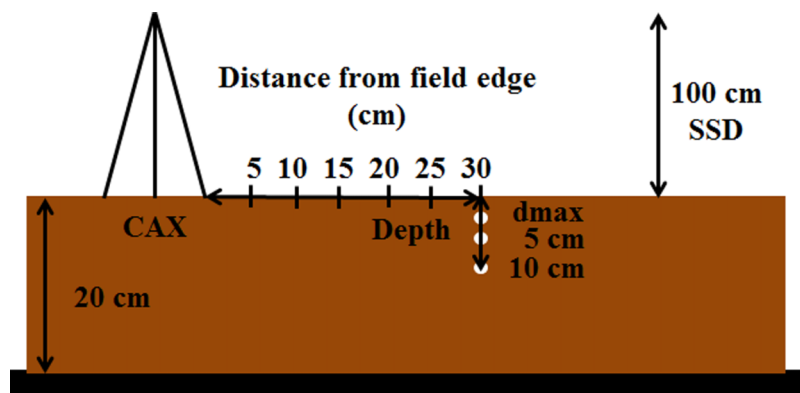


FIG. 1. Schematic of the $40 \times 120 \times 20 \text{ cm}^3$ (width \times length \times depth) solid water phantom used for the PD measurements. The distances from the field edge were defined at a 100 cm source-to-surface distance (SSD). The schematic shows the three depths (nominal depth of maximum dose, d_{max} , 5, and 10 cm) at which the ionization chamber was positioned in the solid water, for example, a 30 cm distance from the field edge.

and the Acuros XB (AXB) algorithm version 10.0.42 with a 0.25 cm grid size. The calculated PD was normalized to CAX at a depth of d_{max} to compare the normalized PD of both AAA and Acuros to the measured data.

3. RESULTS

The trends in the measured PDs for the 6 FFF and 10 FFF photon beams are shown in Fig. 2. The PD for all field sizes (3×3 to 30×30 cm²) and distances (5–40 cm) are shown in Table I for the 6 FFF beam and Table II for the 10 FFF beam.

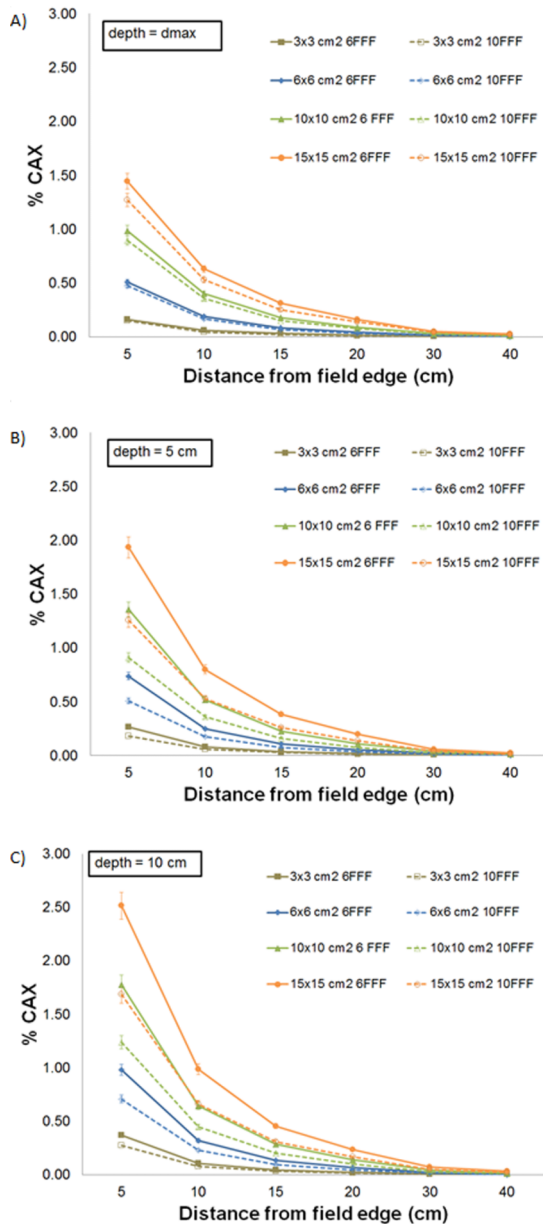


FIG. 2. Measured PDs normalized to the CAX dose at a depth of nominal d_{max} for the 6 FFF (solid line) and 10 FFF (dashed line) beams. The PD measurements are shown at a depth of (A) nominal d_{max} (1.5 cm for the 6 FFF beam and 2.0 cm for the 10 FFF beam), (B) 5 cm, and (C) 10 cm and were acquired for field sizes ranging from 3×3 to 30×30 cm², and distances of 5–40 cm from the field edge with the MLC shaping the field apertures and with a collimator rotation of 90°.

The measured PD at the nominal depth of d_{max} and at a distance of 5 cm from the field edge for a 6×6 cm² square field was 0.51% and 0.48% for the 6 FFF and 10 FFF photon beams, respectively. This compares to the measured PD for a 30×30 cm² field at the same depth and distance from the field edge of 2.01% and 1.63% for the 6 FFF and 10 FFF photon beams, respectively. The measured PD for the 6×6 cm² square field at a distance of 5 cm from the field edge was found to increase from 0.51% to 0.98% and 0.48% to 0.71% when the depth was varied between d_{max} and 10 cm, for the 6 FFF and 10 FFF photon beams, respectively.

The effect of a 0° versus 90° collimator rotation, as well as the impact of a tertiary MLC was also investigated for a 10×10 cm² field size. The difference in PD at depths of d_{max} , 5, and 10 cm for both 0° and 90° collimator rotations and MLC shaped field are shown in Fig. 3 for both the 6 FFF and 10 FFF beams. At a distance of 5 cm from the field edge and a depth of 1.5 cm, the measured PD for the 6 FFF beam was 1.31% versus 1.22% for the 0° and 90° collimator rotations. At the same distance from the field edge and a depth of 2 cm, the measured PD for the 10 FFF beam was 1.32% versus 1.15% for the 0° and 90° collimator rotations. The PD measured at the nominal d_{max} at 10 cm from the field edge was 0.36% versus 0.41% for the 6 FFF beam and 0.29% versus 0.36% for the 10 FFF beam for the 0° and 90° collimator rotations, respectively.

PD measurements for the 6 FFF beam were also compared to the previously acquired PD measurements of a 6 MV beam on a Varian Trilogy linac (Varian Medical Systems, Palo Alto, CA).²⁶ Figure 4 shows the measured PDs normalized to the CAX at a depth of 1.5 cm for the 6 FFF and 6 MV beams which were acquired at depths of 1.5, 5, 10 cm. The PD measurements are shown for field sizes ranging from 6×6 to 30×30 cm², and distances of 5–40 cm from the field edge with the MLC shaping the field apertures and with a collimator rotation of 90°. For larger field sizes (20×20 to 30×30 cm²), the measured PD was smaller for the 6 FFF beam compared to the 6 MV beam, 2.01% compared to 2.57% for the 6 FFF and 6 MV photon beams, respectively, at a distance of 5 cm from the field edge for a 30×30 cm² square field. At a depth of 10 cm, the PDs for the 6 FFF and 6 MV photon beams are nearly identical for the 6×6 and 10×10 cm² fields for all distances from the field edge. Similar results were seen for the 15×15 cm² field for all distances from the field edge except 5 cm, where the measured PD is 2.51% compared to 2.96% for the 6 FFF and 6 MV beams, respectively. For the larger field sizes (20×20 to 30×30 cm²), the measured PD was smaller for the 6 FFF beam compared to the 6 MV beam, 3.52% compared to 5.56% for the 6 FFF and 6 MV photon beams, respectively, at a distance of 5 cm from the field edge for a 30×30 cm² square field.

Peripheral dose was also modeled in ECLIPSE (version 10) by using both the AAA and Acuros dose calculation models. Table III shows the percent error of the measured to calculated PD for the 6 FFF and 10 FFF beams at 3×3 cm² and 6×6 cm² fields at depths of d_{max} , 5, and 10 cm. The difference in measured versus calculated dose has not been normalized to

TABLE I. Percent ratio of the measured PD to the CAX dose at a depth of 1.5 cm for varying field sizes and depths at distances ranging from 5 to 40 cm from the field edge for a 6 FFF photon beam.

Field size (cm)	Depth (cm)	MLC status	Collimator angle (deg)	Distance from the field edge (cm)					
				5	10	15	20	30	40
3 × 3	1.5	Shaped	90	0.16	0.06	0.03	0.02	0.01	0.01
3 × 3	5	Shaped	90	0.26	0.08	0.04	0.02	0.01	0.01
3 × 3	10	Shaped	90	0.37	0.11	0.05	0.02	0.01	0.01
6 × 6	1.5	Shaped	90	0.51	0.19	0.08	0.05	0.02	0.01
6 × 6	5	Shaped	90	0.74	0.25	0.11	0.05	0.02	0.01
6 × 6	10	Shaped	90	0.98	0.32	0.13	0.06	0.23	0.00
10 × 10	1.5	Retracted	0	1.31	0.65	0.41	0.23	0.09	0.04
10 × 10	5	Retracted	0	1.62	0.71	0.41	0.22	0.10	0.04
10 × 10	10	Retracted	0	1.98	0.79	0.42	0.22	0.09	0.04
10 × 10	1.5	Retracted	90	1.22	0.61	0.34	0.15	0.07	0.03
10 × 10	5	Retracted	90	1.56	0.69	0.37	0.17	0.07	0.03
10 × 10	10	Retracted	90	1.96	0.78	0.40	0.18	0.07	0.03
10 × 10	1.5	Shaped	0	0.96	0.36	0.18	0.10	0.04	0.02
10 × 10	5	Shaped	0	1.33	0.48	0.22	0.12	0.04	0.02
10 × 10	10	Shaped	0	1.75	0.61	0.27	0.14	0.04	0.02
10 × 10	1.5	Shaped	90	0.99	0.41	0.18	0.09	0.04	0.02
10 × 10	5	Shaped	90	1.36	0.52	0.23	0.11	0.04	0.02
10 × 10	10	Shaped	90	1.78	0.64	0.28	0.14	0.04	0.02
15 × 15	1.5	Shaped	90	1.45	0.63	0.31	0.16	0.05	0.03
15 × 15	5	Shaped	90	1.93	0.80	0.38	0.20	0.06	0.03
15 × 15	10	Shaped	90	2.51	0.99	0.45	0.23	0.07	0.03
20 × 20	1.5	Shaped	90	1.74	0.79	0.39	0.23	0.07	0.03
20 × 20	5	Shaped	90	2.32	1.01	0.49	0.27	0.09	0.03
20 × 20	10	Shaped	90	3.01	1.23	0.58	0.30	0.10	0.04
25 × 25	1.5	Shaped	90	1.89	0.88	0.44	0.25	0.10	0.04
25 × 25	5	Shaped	90	2.54	1.13	0.56	0.30	0.11	0.05
25 × 25	10	Shaped	90	3.31	1.39	0.67	0.35	0.12	0.05
30 × 30	1.5	Shaped	90	2.01	0.95	0.48	0.28	0.12	0.05
30 × 30	5	Shaped	90	2.69	1.22	0.61	0.34	0.13	0.06
30 × 30	10	Shaped	90	3.52	1.50	0.74	0.39	0.13	0.06

CAX dose and is therefore representative of the local rather than the global error for AAA and Acuros. Between 5 and 10 cm from the field edge, the average percent error was found to be 9% and 9.3% for AAA and Acuros, respectively, at all energies and depths for 3 × 3 cm² and 6 × 6 cm² fields. For the field sizes (3 × 3 to 30 × 30 cm²) evaluated, the average percent error was -13.6% for AAA and -7.8% for Acuros for all depths and distances of 5 and 10 cm from the field edge. At distances from the field edge of 15 cm and greater, AAA reported zero dose for all depths and field sizes. Acuros also reported zero dose at distances from the field edge of ≥15 cm for field sizes ≥10 × 10 cm².

4. DISCUSSION

To demonstrate how this dataset can be implemented clinically, the following simple example is provided. A pa-

tient with a pacemaker is scheduled to undergo radiotherapy for a lung lesion with an average field size of 6 × 6 cm² and a 6 FFF beam. The prescription dose of 60 Gy is to be delivered using a six-field treatment plan that includes anterior–posterior (AP), posterior–anterior (PA), and four oblique treatment beams. A detailed estimate of PD to the pacemaker is requested by the treating physician. For each field, the following methodology is applied: (1) the dose at 1.5 cm depth along the center of the field is found from the treatment planning system, (2) the distance from the beam edge to the most proximal portion of the pacemaker is measured in the treatment planning system, (3) the depth of the pacemaker from the body surface at the closest edge of the treatment field is measured, and (4) the appropriate figures or tables are used to estimate the out-of-field dose to the device. Pacemaker depths less than 1.5 cm are treated as though they were at 1.5 cm, while depths greater than 10 cm are treated as though they were at 10 cm. These assumptions

TABLE II. Percent ratio of the measured PD to the CAX dose at a depth of 2 cm for varying field sizes and depths at distances ranging from 5 to 40 cm from the field edge for a 10 FFF photon beam.

Field size (cm)	Depth (cm)	MLC status	Collimator angle (deg)	Distance from the field edge (cm)					
				5	10	15	20	30	40
3 × 3	2	Shaped	90	0.15	0.05	0.02	0.01	0.01	0.01
3 × 3	5	Shaped	90	0.18	0.06	0.03	0.01	0.01	0.01
3 × 3	10	Shaped	90	0.27	0.08	0.03	0.02	0.01	0.01
6 × 6	2	Shaped	90	0.48	0.17	0.07	0.03	0.02	0.00
6 × 6	5	Shaped	90	0.51	0.18	0.08	0.04	0.01	0.00
6 × 6	10	Shaped	90	0.71	0.23	0.09	0.04	0.02	0.00
10 × 10	2	Retracted	0	1.32	0.66	0.40	0.23	0.09	0.03
10 × 10	5	Retracted	0	1.17	0.54	0.33	0.18	0.08	0.03
10 × 10	10	Retracted	0	1.44	0.59	0.33	0.17	0.08	0.03
10 × 10	2	Retracted	90	1.15	0.55	0.32	0.14	0.06	0.03
10 × 10	5	Retracted	90	1.11	0.52	0.30	0.13	0.05	0.02
10 × 10	10	Retracted	90	1.41	0.58	0.31	0.13	0.05	0.02
10 × 10	2	Shaped	0	0.87	0.29	0.14	0.08	0.03	0.02
10 × 10	5	Shaped	0	0.90	0.32	0.15	0.08	0.03	0.02
10 × 10	10	Shaped	0	1.22	0.42	0.19	0.10	0.03	0.02
10 × 10	2	Shaped	90	0.90	0.36	0.15	0.08	0.03	0.01
10 × 10	5	Shaped	90	0.91	0.36	0.16	0.08	0.03	0.01
10 × 10	10	Shaped	90	1.24	0.45	0.20	0.10	0.03	0.01
15 × 15	2	Shaped	90	1.27	0.53	0.25	0.14	0.04	0.02
15 × 15	5	Shaped	90	1.26	0.53	0.26	0.14	0.04	0.02
15 × 15	10	Shaped	90	1.68	0.66	0.31	0.16	0.05	0.02
20 × 20	2	Shaped	90	1.49	0.64	0.31	0.18	0.06	0.03
20 × 20	5	Shaped	90	1.46	0.65	0.32	0.18	0.06	0.03
20 × 20	10	Shaped	90	1.95	0.80	0.38	0.20	0.07	0.03
25 × 25	2	Shaped	90	1.57	0.68	0.33	0.19	0.08	0.03
25 × 25	5	Shaped	90	1.56	0.71	0.36	0.20	0.08	0.03
25 × 25	10	Shaped	90	2.10	0.88	0.43	0.23	0.08	0.04
30 × 30	2	Shaped	90	1.63	0.72	0.35	0.21	0.09	0.04
30 × 30	5	Shaped	90	1.63	0.75	0.39	0.22	0.09	0.04
30 × 30	10	Shaped	90	2.18	0.93	0.46	0.25	0.09	0.05

are consistent with the references for the out-of-field dose from flattened beams and provide a conservative estimate. As an illustration, an estimate of PD is presented for only the AP field of the six-field plan. The AP field size is $6 \times 6 \text{ cm}^2$ with a 90° collimator rotation delivering a dose of 15 Gy at a 1.5 cm depth along the center of the field. If the pacemaker is 5 cm from the field edge and at a 1.0 cm depth, Table I can be used to estimate the out-of-field dose from the AP beam. A PD of 0.51% is found from the first row of the $6 \times 6 \text{ cm}^2$ field data (indicating a 1.5 cm pacemaker depth) under the column marked “5” (indicating a 5 cm distance from the field edge). The total out-of-field dose from the AP field is calculated as $15 \text{ Gy} \times 0.0051 = 7.7 \text{ cGy}$. Each of the fields can be individually considered in this manner, the result summed, and then reported to the requesting physician.

The PD for both the 6 FFF (see Table I) and 10 FFF (see Table II) photon beams were found to decrease with increasing distance from the radiation field edge and

decreasing field size. Additionally, when comparing the PD measurements for the 6 FFF and 10 FFF beams, the PD was observed to decrease with increasing energy. Overall, the measured PD was observed to be higher for the 6 FFF than for the 10 FFF for all field sizes and depths. These differences became more pronounced with increasing depth and field sizes and for measurement distances closest to the field edge. All PD values were acquired using static MLCs; therefore, use caution when estimating PD for the modulated fields.

When the field aperture was set by the collimator jaws alone, the PD was observed to be moderately higher for a 0° collimator rotation relative to 90° for both the 6 FFF and 10 FFF photon beams. Conversely, when the field aperture is shaped by the MLC, the PDs measured at 0° and 90° are nearly identical for both the 6 FFF and 10 FFF beams with the exception of measurements acquired at all depths for a distance of 10 cm from the field edge. At this distance from

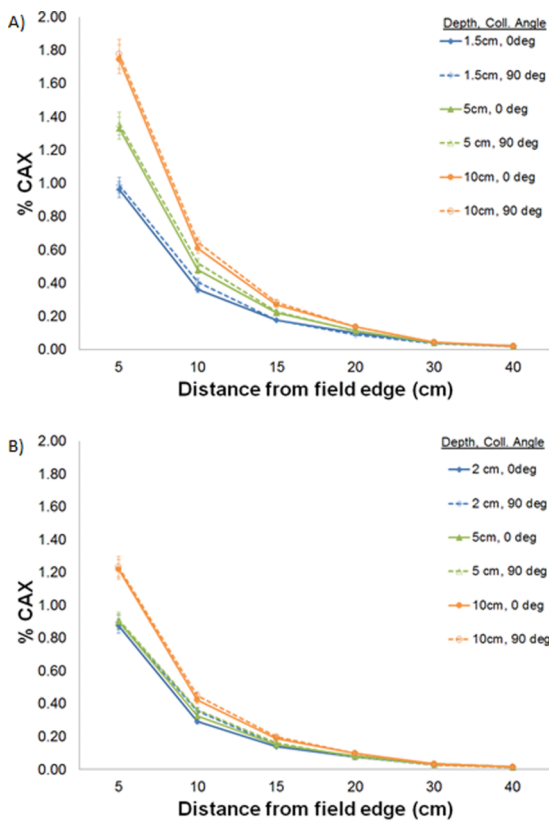


FIG. 3. To evaluate the effects of collimator rotation (0° —solid line and 90° —dashed line) in the presence of the MLC, the measured PDs normalized to the CAX dose at a depth of d_{max} are shown for (A) 6 FFF and (B) 10 FFF photon beams at depths of d_{max} , 5, and 10 cm for both beam energies.

the field edge, the PD was observed to be slightly higher with a 90° collimator rotation. For the Truebeam, a minimal difference in PD was observed between the two collimator rotation angles at distances between 5 and 30 cm from the field edge which is consistent with the 6 MV flattened beam results reported by Mutic and Klein.²⁰ The impact of collimator rotation on PD may be dependent on machine make, model, and beam energy used; therefore, these results should not be assumed to translate across all the machines.²⁸

When comparing the measured PD for the 6 FFF and 6 MV flattened photon beams, the relative PD was found to depend on the depth, field size, and distance from the field edge (see Fig. 4). At a depth of 1.5 cm, the PD for the 6 FFF beam was marginally higher than the 6 MV beam for the 6×6 and 10×10 cm² field sizes up to a distance of 20 cm from the field edge, measuring 0.51% versus 0.46% for the 6 FFF and 6 MV photon beams, respectively, for a 6×6 cm² open field. The PDs of the 6 FFF and 6 MV beam were nearly identical for the 15×15 cm² field size at all measured distances from the field edge.

Since PD is a relative measurement, many uncertainties in the data (e.g., calibration factor) do not impact the accuracy of the values. Since the data were collected over several weeks, the largest source of error is expected to be the uncertainty of the setup relative to previous measurements. To ameliorate this uncertainty, repeat measurements were

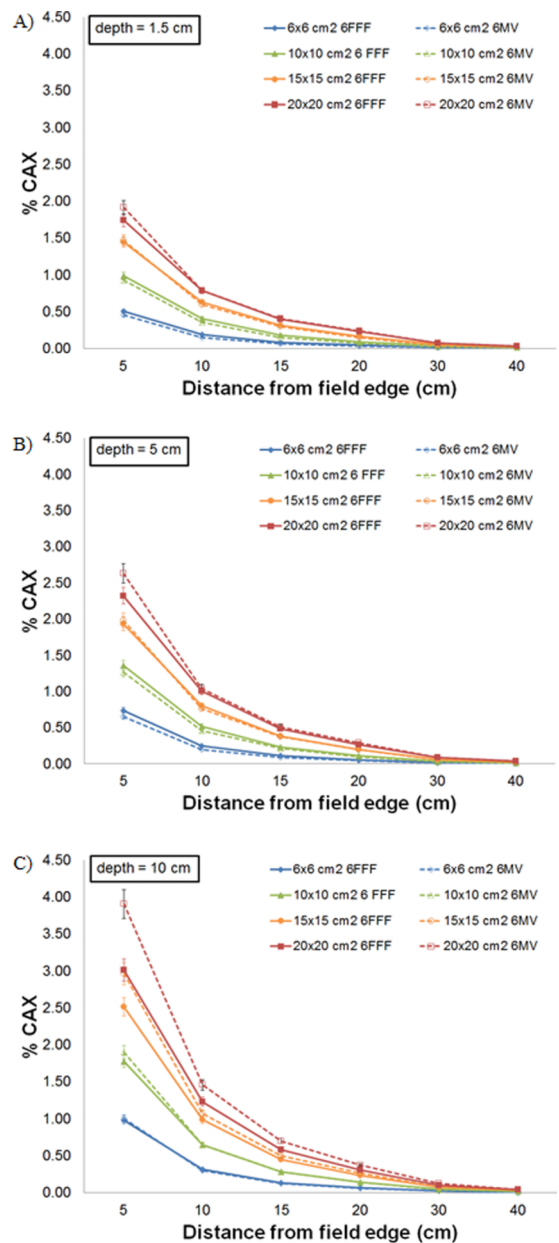


FIG. 4. Measured PDs normalized to the CAX dose at a depth of 1.5 cm for the 6 FFF (solid line) and 6 MV (dashed line) beams acquired at depths of (A) 1.5 cm, (B) 5 cm, and (C) 10 cm. The PD measurements are shown for field sizes ranging from 6×6 to 30×30 cm², and distances of 5–40 cm from the field edge with the MLC shaping the field apertures and with a collimator rotation of 90° .

performed over the various data collection sessions to verify reproducibility of the set-up as well as the stability of equipment (e.g., ion chamber, electrometer). Please note, a limitation of the current study is that a single measurement was acquired for many data points due to the large number of measurements (greater than 400) required to monitor the PD trends. As such, standard deviations were unable to be calculated. Repeat measurements were performed on a subset of the data to address reproducibility. From nearly 30 repeat measurements of various energies, field sizes, and distances from the field edge, the average deviation from the initial measurement was less than 4%. However, when PD is on

TABLE III. Percent error of measured PD to calculated PD for both the AAA and Acuros dose calculation models for 3×3 and 6×6 cm² fields.

Energy (MV)	Field size (cm ²)	Depth (cm)	% error from measurement					
			5 (cm)		10 (cm)		15 (cm)	
			AAA	Acuros	AAA	Acuros	AAA	Acuros
6 FFF	3×3	1.5	-1.2	-6.1	-30.5	-13.6	—	-68.8
		5	3.8	-6.8	-13.4	-12.2	—	-44.4
		10	7.3	-5.7	-2.9	-12.4	—	-32.6
	6×6	1.5	-11.4	-12.4	-27.9	-6.3	—	-28.0
		5	1.9	-1.1	-13.7	-8.8	—	-18.3
		10	10.8	3.0	-4.4	-0.3	—	-9.2
10 FFF	3×3	2	-10.0	-9.3	0.0	0.0	—	-50.0
		5	-8.2	-13.6	-19.6	-19.6	—	-60.7
		10	3.3	-8.4	-12.8	-26.9	—	-30.3
	6×6	2	-24.6	-12.4	-28.3	-7.8	—	-33.3
		5	-6.3	-9.6	-20.3	-19.8	—	-28.6
		10	2.7	-4.1	-9.3	-8.4	—	-18.1

the order of 0.01% or less of the CAX dose, the uncertainty is dominated by the resolution of the electrometer. Pion for the ion chamber used was 1.012 at central axis under calibration conditions for the 10 FFF beam. Compared to a conventional flattened beam, this represents an approximately 0.7% increase in Pion that could indeed contribute to the measured error when the values are normalized to central axis. Combining this error in quadrature with setup error results in a combined error of 4.1%; therefore, we have conservatively added 5% error bars to all data points.

Similar to previous studies,^{18,25} the PD calculated using both AAA and Acuros was typically found to underestimate the dose when compared to the measured values. The difference in measured and calculated dose typically increased as the distance from the field edge increased for both the algorithms although exceptions were noted. For example, the percent error in the TPS calculated dose for a 10 FFF 6×6 cm² field at 5 cm from the field edge was -24.6%. The percent error decreased to -7.8% at 10 cm from the field edge. For small field sizes, where FFF beams are primarily advantageous (e.g., SBRT and SRS), the underestimation of the PD dose with AAA and Acuros was of a similar magnitude.

Comparisons between the commercial algorithms and measurements at distances greater than 10 cm from the field edge were not performed, since AAA reports zero dose for the designated points of interest that were beyond 10 cm from the field edge. AAA uses a divergent dose matrix where the width of the calculation matrix is dependent on the jaw position. The default margin is 12 cm from the field but can range for 7–12 cm to limit the number of calculation points. Acuros uses the same margins for input fluences, but extends the entire dose calculation to the entire calculation volume [Varian Medical Systems (2014). Eclipse Photon and Electron Reference Guide]. Even so, at distance of 15 cm and greater, Acuros only reported calculated doses up to field sizes of 10×10 cm². Acuros also underestimated the

PD at 15 cm from the field edge for both the 6 FFF and 10 FFF beams at field sizes of 3×3 and 6×6 cm² with an average percent error of -35%. The maximum percent error (-68.8%) was observed with a 6 FFF energy at 15 cm from a 3×3 cm² field at 1.5 cm depth. This level of uncertainty in calculated dose has clinical significance for estimating dose to out-of-field organs at risk (OARs). For example, when estimating the dose to a fetus using a 3×3 cm² field at 1.5 cm depth and 15 cm from the field edge, Acuros reports PD of 0.01%. For a 50 Gy treatment, the estimated dose to the fetus is 5 mGy. When using Table I, PD was measured to 0.03% and the estimated dose would be 15 mGy. The average percent error from the measured PD compared to Acuros for field sizes 3×3 , 6×6 , and 10×10 cm² at 15 cm from the field edge was -29%. It is highly recommended that TPS calculated PD should be verified by measurement when making critical clinical decisions on out-of-field OARs during treatment planning.

The Varian TrueBeam representative beam data used for the models incorporate profiles that typically extend 5 cm beyond the nominal field edge. This dataset was selected because it is widely available and frequently encountered in the clinical setting. It may be possible to improve agreement with measured doses by using a commissioning dataset that extends further outside of the field edge and developing a beam model that uses parameters optimized for extrafocal radiation components. The manufacturer (Varian Medical Systems, Palo Alto, CA) was not able to clarify how much improvement could be expected if such an approach was used.

Neutron dose is also of concern when considering peripheral dose to electronic implantable devices, pediatric patients or fetus. The energies used in this paper do not exceed 10 MV; therefore, neutron dose is not expected contribute significantly to PD. In FFF beams, the absence of the flattening filter and reduced photon fluence are expected to reduce neutron contamination. We refer readers to the AAPM

report on FFF beams for a detailed discussion on FFF beams and neutron dose.¹⁰

5. CONCLUSIONS

PD was measured for various field sizes (3×3 to 30×30 cm²) and distances from field edge (5–20 cm) for the 6 FFF and 10 FFF beams in a solid water phantom. This comprehensive dataset can be used to estimate PD to out-of-field regions of interest such as organs at risk, electronic implantable devices, and the fetus. Peripheral dose was found to decrease with increasing energy and increase with increasing depth in the phantom. The impact of collimator rotation (0° versus 90°) was found to slightly decrease PD; however, when combined with tertiary MLCs, the impact of collimator rotation was not clinically significant. Peripheral dose was calculated in a commercial treatment planning system by using both AAA and Acuros dose calculation algorithms. Both algorithms underestimated the PD, and the percent error between measured and calculated PD varied with field size and distance from the field edge; therefore, physicists should use caution when interpreting and applying TPS calculated PD to determine dose to critical structures.

CONFLICT OF INTEREST DISCLOSURE

The authors have no COI to report.

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