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Abstract

Purpose: Flattening filter free (FFF) beams are increasingly being considered for stereotactic radiotherapy (SRT). For the first time, the performance of a monolithic silicon array detector under 6 and 10 MV FFF beams was evaluated. The dosimeter, named "Octa" and designed by the Centre for Medical Radiation Physics (CMRP), was tested also under flattened beams for comparison. **Methods:** Output factors (OFs), percentage depth-dose (PDD), dose profiles (DPs) and dose per pulse (DPP) dependence were investigated. Results were benchmarked against commercially available detectors for small field dosimetry. **Results:** The dosimeter was shown to be a 'correction-free' silicon array detector for OFs and PDD measurements for all the beam qualities investigated. Measured OFs were accurate within 3% and PDD values within 2% compared against the benchmarks. Cross-plane, in-plane and diagonal DPs were measured simultaneously with high spatial resolution (0.3 mm) and real time read-out. A DPP dependence (24% at 0.021 mGy/pulse relative to 0.278 mGy/pulse) was found and could be easily corrected for in the case of machine specific quality assurance applications. **Conclusions:** Results were consistent with those for monolithic silicon array detectors designed by the CMRP and previously characterized under flattened beams only, supporting the robustness of this technology for relative dosimetry for a wide range of beam qualities and dose per pulses. In contrast to its predecessors, the design of the Octa offers an exhaustive high-resolution 2D dose map characterization, making it a unique real-time radiation detector for small field dosimetry for field sizes up to 3 cm side.

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A novel high-resolution 2D silicon array detector for small field dosimetry with FFF photon beams

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Purpose: Flattening filter free (FFF) beams are increasingly being considered for stereotactic radiotherapy (SRT). For the first time, the performance of a monolithic silicon array detector under 6 and 10 MV FFF beams was evaluated. The dosimeter, named “Octa” and designed by the Centre for Medical Radiation Physics (CMRP), was tested also under flattened beams for comparison.

Methods: Output factors (OFs), percentage depth-dose (PDD), dose profiles (DPs) and dose per pulse (DPP) dependence were investigated. Results were benchmarked against commercially available detectors for small field dosimetry.

Results: The dosimeter was shown to be a ‘correction-free’ silicon array detector for OFs and PDD measurements for all the beam qualities investigated. Measured OFs were accurate within 3% and PDD values within 2% compared against the benchmarks. Cross-plane, in-plane and diagonal DPs were measured simultaneously with high spatial resolution (0.3 mm) and real time read-out. A DPP dependence (24% at 0.021 mGy/pulse relative to 0.278 mGy/pulse) was found and could be easily corrected for in the case of machine specific quality assurance applications.

Conclusions: Results were consistent with those for monolithic silicon array detectors designed by the CMRP and previously characterized under flattened beams only, supporting the robustness of this technology for relative dosimetry for a wide range of beam qualities and dose per pulses. In contrast to its predecessors, the design of the Octa offers an exhaustive high-resolution 2D dose map characterization, making it a unique real-time radiation detector for small field dosimetry for field sizes up to 3 cm side.

Key words: SRT, SBRT, FFF, small field dosimetry QA, 2D monolithic silicon array detector

1. Introduction

Stereotactic radiotherapy (SRT) techniques, of which stereotactic body radiation therapy (SBRT) is an example, are a form of external beam radiotherapy (EBRT). These treatments deliver high doses in just a few fractions, up to 45 Gy/fraction in the case of SBRT, using small radiation fields^{1,2}.

Codes of Practice for quality assurance (QA) in the case of small field dosimetry have been only recently outlined^{3,4}. Challenges associated with this scenario are beam related, such as partial occlusion of the primary source and loss of CPE on the central axis, and detector

related, relative to its dimensions with respect to the field and its perturbation effects on the particles spectra^{3, 4}. These conditions, resulting in overlapping penumbrae over the detector volume, may affect its readings, thus the accuracy of the treatment planning system (TPS) in predicting dose distributions. Dosimetric inaccuracies may lead to poor outcomes for patients^{1, 3}.

Recently, a growing interest in rapid delivery of heterogeneous dose distributions has revived the use of flattening filter free (FFF) beams⁵. The removal of the flattening filter from the LINAC changes the profile and dosimetric characteristics of radiation beams⁶. Reported clinical benefits are mainly a result of an increased available dose rate and lower peripheral doses (PD)⁷. With higher dose per pulses and dose profiles having steeper gradients, FFF beams compound all the problems associated with small field dosimetry for flattened beams and may prove challenging for dosimeters performance^{6, 7}.

Dosimeters for SRT QA are to be water equivalent, dose-rate independent, with a good signal to noise ratio and real-time read-out^{3, 4}. They should have a sufficiently small sensitive volume to avoid volume-averaging effects³, which are related to the dose gradients over the sensitive volume⁵ and can result in a different signal compared to the signal a point-like detector would measure. To date, in the absence of such an ideal dosimeter, it has been common practice to perform QA measurements with at least two types of radiation detectors and then crosscheck the results for consistency⁸, often along with the use of Monte Carlo (MC) simulations. Several alternatives have been described in the literature.

EBT3 Gafchromic films have minimal energy dependence and offer high spatial resolution but not real-time readings, which are also affected by large uncertainties due to film polarization, non-uniformity, scanning and handling techniques⁸. Ionization chambers (IC) are the recognized standard for large field dosimetry but are impaired by volume-averaging effects when used for small radiation fields⁴. Diamond-based detectors have been employed for routine QA thanks to their water equivalence, energy independence and high signal to noise ratio^{3, 4}, but are expensive and as such not widely employed. Furthermore, they exhibit dose rate dependence, though corrections can be applied³. All these dosimeters are subject to central axis (CAX) alignment problems, an issue all the more relevant for small radiation fields⁴.

Silicon diodes are a valuable option for small field dosimetry thanks to their large dynamic range and high sensitivity, real-time operations, well-developed manufacturing technology and high spatial resolution due to the small sensitive volumes (SVs). However, they are known to be dose rate dependent, with an increase in sensitivity with dose per pulse reported for p-type silicon diodes^{9, 10}.

Furthermore, correction factors need to be applied to account for beam perturbations, due to their SVs and extra-cameral components. These factors depend on detector design, treatment head design, beam quality, field size and measurement conditions³.

It was shown that it is possible to design a 'correction-free' detector, though, with the addition of low density media to the high density detector components¹¹. However, it must be verified that these modifications are correctly compensating whatever the beam quality and measurements conditions¹².

2D monolithic silicon diode array detectors, with either 2 mm and 3 mm pitch, have been shown to be promising as dosimeters by several groups^{13, 14}. Commercially available options based on single diodes are the ArcCHECK (Sun Nuclear Corp., Melbourne, FL) and the Delta4 (ScandiDos AB, Uppsala, Sweden). Their spatial resolution, though, is not adequate

for small field dosimetry. In fact, while with 1D monolithic detectors it is easy to decrease the pitch between silicon diodes down to 0.2 mm (CMRP DMG) ¹⁵, in the case of 2D detectors a compromise is necessary between the overall active area and the spatial resolution provided, in order to be within limitations in the number of read-out channels.

The Centre for Medical Radiation Physics (CMRP), University of Wollongong, has designed and characterized two 1st generation monolithic silicon diode array detectors for SRT QA, the MP512 ¹⁶ and the Duo ¹⁷. In those studies, they were shown to be accurate dosimeters for output factors (OFs), percentage depth dose (PDD) and dose profile measurements under flattened beams with a dose per pulse (DPP) dependence. The angular dependence of the MP512 was investigated and could be corrected for, making it a suitable candidate for arc therapy delivery QA ¹⁸. The rather coarse spatial resolution (2 mm) of the MP512 and the limited characterization of the 2D dose map given by the Duo, though, impair their attractiveness for contemporary small field dosimetry where sub-millimetre spatial resolution and a detailed description of the 2D dose map is paramount, especially when using FFF beams.

The Octa, a 2nd generation monolithic silicon diode array detector, incorporates its predecessors' technology and as such, it is characterized by the same signal stability, radiation hardness and dose linearity. The Octa's 512 diodes-SVs are arranged in four intersecting orthogonal linear arrays such that cross-plane, in-plane and 2 diagonal dose profiles are characterized simultaneously with sub-millimetre resolution.

This study evaluated the potential of the Octa for relative dose measurements, in particular in the challenging measurements conditions of small fields with FFF beams. Parameters commonly used by commercial TPSs, such as dose profiles, PDD curves and OFs were investigated. Results were benchmarked against those for other commercially available dosimeters. In order to have a comprehensive analysis of the Octa performance, 6 and 10 MV flattened beams were included in the study.

2. Materials and methods

2.1. The dosimeter

The Octa, pictured in Figure 1, is a 2D monolithic silicon array detector based on SVs fabricated on a high resistivity p-type epitaxial ¹⁹, grown on top of a low resistivity p⁺ substrate. A thin protective layer of epoxy covers the SVs. The 512 diodes have all the same sensitive area of 0.032 mm² and are of elongated rectangular shape (0.04 mm x 0.8 mm), except for the 9 pixels in the central matrix at the intersection of the 4 arrays (0.160 mm x 0.200 mm). The device has a sub-millimetre resolution with diodes having a 0.3 mm pitch along the vertical and horizontal arrays and a 0.43 mm pitch along the 2 diagonal arrays. The diodes are operated in passive mode, i.e. with no bias voltage applied, and connected to a multichannel readout electronics data acquisition (DAQ) system based on a commercially available analogue front end (AFE0064, Texas Instruments), which was previously described in detail ^{16, 20}. In this study, an equalization procedure ²¹ was used to correct for small differences in each channel response. This variability is due to a small difference in the sensitivity of each diode and the gain of its corresponding preamplifier in an application-specific integrated-circuit (ASIC).

Conceived as a 2D planar dosimeter for dose measurements in solid water, the Octa is sandwiched between two Perspex plates, each 5 mm thick. A small air gap on top of its SVs

minimizes the number and size of corrections that are required to relate its readings to dose²².

Experimental measurements described in this study were carried out at the Illawarra Cancer Care Centre, Wollongong NSW, Australia using a Varian Clinac LINAC (Varian Medical Systems, Palo Alto CA) and at the Peter MacCallum Cancer Centre, Melbourne VIC, Australia using a Varian TrueBeam STx LINAC (Varian Medical Systems, Palo Alto CA), as summarized in Table 1. Dosimeters used as reference are summarized in Table 2.

Figure 1. The Octa is a 2D monolithic silicon array detector consisting of 512 diodes operated in passive mode and arranged in four intersecting orthogonal linear arrays. Each diode has a sensitive area of 0.032 mm^2 with pitch of 0.3 mm along the vertical and horizontal arrays and of 0.43 mm along the 2 diagonal arrays.

2.3 Output factors and dose profiles

Output factors are defined as the ratio between the detector reading at specific field size (*clin*) and that at the machine specific reference field (*msr*), following the formalism used by Francescon et al.²³:

$$OF_{det} = \frac{M^{f_{clin}}}{M^{f_{msr}}}$$

where $M^{f_{clin}}$ and $M^{f_{msr}}$ are the corrected detector readings in the f_{clin} and f_{msr} fields respectively.

The OFs were measured at 90 cm source to surface distance (SSD) and 10 cm depth in solid water. Dose profiles were acquired at 90 cm SSD and 10 cm depth in solid water for measurements at the Illawarra Cancer Care Centre, and at 100 cm SSD and 10 cm depth for measurements at the Peter MacCallum Cancer Centre, in order to follow the QA protocol in place.

Prior to measurements, the Octa was aligned with respect to the CAX by maximizing the response of its central pixel using the smallest available field size, a 5 mm side square field.

For OFs measurements, the detector reading at each field size was taken as the average response of its central pixel over 3 repetitions of the same measure followed by normalization of these averages to the average reading at the reference field size.

For dose profiles, the Octa reading at each field size was taken as the reading of each channel averaged over 3 repetitions of the same measure followed by normalisation of the response of each channel to the median response of the pixels within 0.5 mm of CAX.

Dose profiles were evaluated by comparing FWHM and penumbra values, which was taken as the distance between the 80% and the 20% isodose levels. For a quantitative estimation, profiles were analysed with MATLAB (Mathworks, Inc.) using a shape preserving interpolant function.

2.4 Dose per pulse dependence

The DPP dependence, which refers to the change of the detector sensitivity due to a change of dose per pulse, was studied by irradiating the Octa with a fixed number of monitor units (MU) and changing the SSD to change the dose per pulse at the detector location^{9, 14}.

The range of doses per pulse investigated was between a maximum of 0.977 mGy/pulse for the 10 MV FFF beam quality and a minimum of 0.021 mGy/pulse for the 6 MV flattened beam quality, as summarized in Table 3. A 10 cm side square field size was used for all SSDs and beam qualities, with measurements were carried out at 1.5 cm depth for the 6 MV flattened beam, but at 10 cm depth for the 6 and 10 FFF beams.

The Octa DPP sensitivity was defined as the ratio of the charge measured by the detector to the charge measured by the ionization chamber used as the reference dosimeter, at the same SSD, i.e. for the same dose per pulse value. The Octa DPP dependence was then taken as its sensitivity at each dose per pulse, normalized to that at the dose per pulse at 100 cm SSD 1.5 cm depth for the 6 MV flattened beam, and to that at 100 cm SSD 10 cm depth for the 6 and 10 MV unflattened beams, as summarized in Table 3.

The two-voltage method, which was deemed accurate in the dose per pulse range investigated²⁴, was used to evaluate the ion recombination correction factor to correct the Farmer ionization chamber readings in the case of 6XFFF and 10XFFF beam qualities. No correction factor was applied to the Farmer ionization chamber readings in the case of 6 MV flattened beam⁶.

2.5 Percentage depth dose

CAX PDD curves were measured with the Octa at 100 cm SSD, with 10 cm solid water for backscattering purposes. A 10 cm side square field size was used for all beam qualities investigated and the desired water depths were reached adding the required amount of solid water slabs on top of the detector.

The average of measurements carried out with a both a CC13 and a Markus ionization chamber under the same experimental conditions was used as reference.

For a quantitative estimation of the percentage differences between the Octa PDDs and those for the reference dosimeters, acquired values were analysed with MATLAB (Mathworks, Inc.) using a shape preserving interpolant function.

2.6 EBT3 films

The EBT3 Gafchromic films used in this study were sandwiched between Perspex plates as used for the Octa. Films were scanned with an EPSON expression 10000XL using a 48-bit RGB with a resolution of 72 dpi. All films were pre- and post-scanned six times using only the last three optical density maps, maintaining a consistent orientation. The film analysis method employed was the same as that used by Aldosari et al.¹⁴.

2.7 Percentage differences between the Octa and reference dosimeters and uncertainty estimation

In all cases, the percentage differences between the Octa readings and those for the dosimeters used as reference was presented as below:

$$\%diff = \frac{Octa - dosimeter_{ref}}{dosimeter_{ref}} \times 100$$

For all measurements, we defined the final reading of each one of the Octa 512 channel as the mean value over 3 repetitions of the same measure with error bars calculated as 2 standard deviations. For the DPP dependence investigation, the error bars shown are the results of the error propagation of the statistical dispersion of both the Octa and the ionization chamber measurements.

Table 1. Participating centres and characteristics of LINACs and beam qualities used. All LINACs were calibrated to deliver 1 cGy/MU at d_{max} in water at 100 cm SSD. 6X and 10X refer to 6 and 10 MV flattened beam, respectively. 6XFFF and 10XFFF refer to unflattened beams.

Centre	Machine	Collimator	Energy [MeV]
Illawarra Cancer Care Centre, Wollongong NSW, Australia	Varian Clinac Varian Medical Systems, Palo Alto CA	jaws	6X
Illawarra Cancer Care Centre, Wollongong NSW, Australia	Varian Clinac Varian Medical Systems, Palo Alto CA	jaws	10X
Peter MacCallum Cancer Centre, Melbourne VIC, Australia	Varian TrueBeam STx Varian Medical Systems, Palo Alto CA	jaws	6XFFF
Peter MacCallum Cancer Centre, Melbourne VIC, Australia	Varian TrueBeam STx Varian Medical Systems, Palo Alto CA	jaws	10XFFF

Table 2. Reference dosimeters used for benchmarking the Octa for this study.

Dosimeter	Comments	Type	Vendor
EBT3 Gafchromic films	EPSON expression 10000XL scanner	dosimetry films	Ashland Inc., USA
microDiamond		synthetic diamond	PTW, Freiburg, Germany
CC13		ionization chamber	IBA Dosimetry GmbH, Germany
Farmer chamber (a)	Type IBA-FC-65P	ionization chamber	IBA Dosimetry GmbH, Germany
Farmer chamber (b) Markus	Type NE2571A Model N23343	ionization chamber ionization chamber	PTW, Freiburg, Germany

Table 3. The Octa DPP dependence investigation: range of DPP used for each beam quality and reference dosimeters. For each beam quality, results were normalized to those for the reference dose per pulse indicated.

Range of dose per pulses investigated [mGy/pulse]	Reference dose per pulse [mGy/pulse]	Size of square field [mm]	Collimator	Beam quality	Reference dosimeter
0.021 to 0.278	0.278	100	jaws	6X	Farmer chamber (b)

0.041 to 0.416	0.416	100	jaws	6XFFF	Farmer chamber (a)
0.079 to 0.977	0.797	100	jaws	10XFFF	Farmer chamber (a)

3. Results

3.1 Output factors

The OFs for the Octa are shown in Figure 2 for the 6 and 10 MV flattened beams, and in Figure 3 for the 6 and 10 MV unflattened beams. In the proposed figures, OFs were normalized to the 30 mm side square field, the smallest available field size for which CPE is still provided for all energies²⁵.

Figure 2. (a) OFs measured by the Octa and EBT3 films for a 6 MV flattened beam for field sizes from 5 mm side square field to 100 mm side square field, normalized to 30 mm side square field. (b) OFs measured by the Octa and EBT3 films for a 10 MV flattened beam for field sizes from 5 mm side square field to 100 mm side square field, normalized to 30 mm side square field. Percentage differences are shown in the lower panels.

Figure 3. (a) OFs measured by the Octa and microDiamond, IC for a 6 MV unflattened beam for field sizes from 5 mm side square field to 100 mm side square field, normalized to 30 mm side square field. (b) OFs measured by the Octa and microdiamond/IC for a 10 MV unflattened beam for field sizes from 5 mm side square field to 100 mm side square field, normalized to 30 mm side square field. Percentage differences are shown in the lower panels. Reference data was acquired using a daisy-chain method with microDiamond (for field sizes equal and smaller than 3 cm side) and ionization chamber (for field sizes equal and larger than 3 cm side), and was not available for the 5 mm side square field, for both beam qualities.

3.2 Dose profiles

Selected dose profiles measured by the Octa are shown in Figure 4 to Figure 9 for all beam qualities investigated and with the corresponding benchmark, where available. Corresponding FWHM and penumbra values for the in-plane and cross-plane profiles are shown in Table 4 and Table 5.

In the figures, profiles were aligned such that the origin lies at the coordinate corresponding to the 50% response.

Figure 4. Dose profiles measured by the Octa and EBT3 films for a 10 MV flattened beam, 5 mm side square field. Profiles are aligned with respect to the 50% peak response, taken as the median value around the CAX.

Figure 5. Dose profiles measured by the Octa and EBT3 films for a 10 MV flattened beam, 10 mm side square field. Profiles are aligned with respect to the 50% peak response, taken as the median value around the CAX.

Figure 6. Dose profiles measured with by Octa and EBT3 films for a 10 MV flattened beam, 30 mm side square field. Profiles are aligned with respect to the 50% peak response, taken as the median value around the CAX.

Figure 7. Dose profiles measured by the Octa and microDiamond for a 6 MV unflattened beam, 10 mm side square field. Profiles are aligned with respect to the 50% peak response, taken as the median value around the CAX. Reference data was not available for diagonal profiles.

Figure 8. Dose profiles measured by the Octa and microDiamond for a 6 MV unflattened beam, 30 mm side square field. Profiles are aligned with respect to the 50% peak response, taken as the median value around the CAX. Reference data was not available for diagonal profiles.

Figure 9. Dose profiles measured by the Octa and microDiamond for a 10 MV unflattened beam, 30 mm square field. Profiles are aligned with respect to the 50% peak response, taken as the median value around the CAX.

Table 4. Summary of FWHM and penumbra values measured by the Octa and the reference dosimeter, for the in-plane profiles presented in Figure 4 to Figure 9.

Beam quality	Square field size [mm]	Octa (in-plane)		Reference (in-plane)		Difference	
		FWHM [mm]	Penumbra [mm]	FWHM [mm]	Penumbra [mm]	Δ FWHM (%)	Δ Penumbra [mm]
10X	5	5.1	2.2	5.4	2.2	-5.6	0.0
10X	10	9.9	3.1	10.0	3.1	-1.0	0.0
10X	30	30.2	4.0	30.4	4.0	-0.7	0.0
6XFFF	10	10.5	3.2	10.2	2.7	2.9	0.5
6XFFF	30	32.5	4.0	32.3	3.4	0.6	0.6
10XFFF	30	32.6	4.9	32.0	3.8	1.9	1.1

Table 5. Summary of FWHM and penumbra values measured by the Octa and the reference dosimeter, for the cross-plane profiles presented in Figure 4 to Figure 9.

Beam quality	Square field size [mm]	Octa (cross-plane)		Reference (cross-plane)		Difference	
		FWHM [mm]	Penumbra [mm]	FWHM [mm]	Penumbra [mm]	Δ FWHM (%)	Δ Penumbra [mm]
10X	5	5.1	1.9	5.2	2.0	-1.9	-0.1
10X	10	10.2	2.7	9.9	2.9	3.0	-0.2
10X	30	29.6	3.9	30.0	3.5	-1.3	0.4
6XFFF	10	9.1	2.6	9.2	2.8	-1.1	-0.2
6XFFF	30	31.3	3.3	31.2	3.4	0.0	-0.1
10XFFF	30	31.1	3.7	31.2	4.4	-0.3	-0.7

3.3 Dose per pulse dependence

The Octa DPP dependence is shown in Figure 10 for all beam qualities investigated.

Figure 10. The Octa response measured against the ionization chamber as a function of dose per pulse. (a) DPP dependence for a 6 MV flattened beam, with ratios normalized to the dose per pulse at 100 cm SSD 1.5 cm depth (0.278 mGy/pulse). (b) DPP dependence for a 6 MV unflattened beam, with ratios normalized to the dose per pulse at 100 cm SSD 10 cm depth (0.416 mGy/pulse). (c) DPP dependence for a 10 MV unflattened beam, with ratios normalized to the dose per pulse at 100 cm SSD 10 cm depth (0.797 mGy/pulse). Error bars represent the combined uncertainties.

3.4 Percentage depth dose

PDD curves measured by the Octa are shown in Figure 11 for the 6 MV flattened beam and in Figure 12 for the 6 and 10 MV unflattened beams. For the Octa, nominal depths were

converted to water equivalent depths to account for the density of the Perspex plates. PDD values for the 6XFFF and 10XFFF beam qualities were measured only up to 10 cm nominal depth in solid water due to limited availability of solid water slabs at the Peter MacCallum Cancer Centre.

Figure 11. (a) PDD measured by the Octa and ionization chamber for a 6 MV flattened beam, 10 cm side square field. Experimental values were analysed using a shape preserving interpolant function. Percentage differences are shown in the lower panel.

Figure 12. (a) PDD measured by the Octa and microDiamond for a 6 MV unflattened beam, 10 cm side square field. (b) PDD measured by the Octa and microDiamond for a 10 MV unflattened beam, 10 cm side square field. Experimental values were analysed using a shape preserving interpolant function. Percentage differences are shown in the lower panels.

4. Discussion

4.1 Output factors

The central pixels of the Octa (0.16 mm x 0.2 mm) were small enough to measure accurately the position of the central axis (CAX) peak without any volume-averaging effect. Once aligned to the CAX, OFs for the Octa were measured for both small and large radiation fields. Positioning uncertainties were therefore negligible, in stark contrast with single detectors for which this is a major source of error in OFs measurements.

Silicon diodes are known to over-respond to photons of low energy because of the increasing cross-section of the photoelectric effect in silicon compared to water²⁶ with the electron density of the extra-cameral components also playing a role. This is exacerbated by the removal of the flattening filter from the LINAC, which results in most of the low-energy photons to pass through and a consequent lower average beam energy⁶. As previously reported, though, deviations in small fields correction factors for silicon diodes between flattened and FFF beams are sufficiently small (up to a maximum of $\pm 1.7\%$) to allow for their potential interchangeability on the same LINAC²⁷. Consistently with this result, we used the same air gap to render the Octa a ‘correction-free’ dosimeter for OFs measurements for all beam qualities investigated.

For all beam qualities investigated, OFs for the Octa were accurate within 3% with respect to values measured with the reference dosimeter.

A Monte Carlo numerical correction factor would be useful for an evaluation of the extra-cameral effect for the Octa and for monolithic silicon array detectors in general, but goes beyond the scope of this work.

4.2 Dose profiles

FWHM values for the Octa for in-plane, cross-plane and diagonal dose profiles were well within 3% with respect to the reference, thus not clinically significant. Exception was the in-plane profile relative to the 5 mm side square field for the 10 MV flattened beam, for which the percentage difference for the FWHM value was found to be 5.6%.

It should be emphasized that small differences between nominal and effective field sizes for small jaws-defined radiation fields, due to the jaws calibration and or their positioning inaccuracies, are known to have a strong impact on small field measurements^{28, 29}. Ideally, dose profiles for both the dosimeter being evaluated and that used as reference would have to be measured at the same time.

4.3 Dose per pulse dependence

Dose per pulse measurements are often difficult as the spectral composition of a beam changes with attenuation and distance from the source (due to contamination). They are a known limitation of silicon-based dosimeters is their dependence on dose per pulse under LINAC irradiation. As first reported by Rikner and Grusell³⁰, a decrease in sensitivity is expected with decreased dose per pulse. While at low dose per pulses the recombination centres near the band edges of the silicon are empty with part of the charge carriers generated by the ionizing radiation being lost to these traps, at high dose per pulses the fraction of these that recombine decreases and a larger portion of the signal is available to be collected⁹.

The Octa was shown to have a DPP dependence in FFF beams comparable to that of other solid-state dosimeters that are considered stable, for the whole range of doses per pulse investigated. A maximum DPP dependence of 24% at 0.021 mGy/pulse, relative to 0.278 mGy/pulse, was found and could be easily corrected for in the case of machine specific QA applications.

At the higher dose per pulse of the 10XFFF beam quality, a difference in the relative response of the pixels in the central matrix of the detector was noted. Since the sensitivity of the diode is proportional to the diffusion length, which is a function of the dose rate, at high dose per pulses there may be an enhanced effect of charge sharing between neighbouring pixels. For the Octa, this effect would be appreciable only for the central pixels due to the perimeter of their sensitive area being more than twice that of the other pixels. Further investigation is in order, but beyond the scope of this work.

4.4 Percentage depth dose

With increasing depth, silicon diodes are expected to overestimate the dose due to the increase of the relative number of low energy scattered photons for clinical photon beams, an effect which could be offset by an underestimation due to dose rate dependence^{31, 32}.

For the Octa, while a DPP dependence was found, discrepancies in PPD with respect to the reference values were within 2% at all depths, for all beam qualities, in a worst-case scenario of a 10 cm side square field.

Due to a limited availability of solid water slabs at Peter MacCallum Cancer Centre, we were unable to measure PDD beyond 10 cm depth. Based on the excellent comparison between the Octa and the ionization chamber PDD curves for the 6 MV flattened beam, though, we don't expect any relevant differences for the unflattened beams.

5. Conclusions

The Octa was demonstrated to be an accurate dosimeter for small field relative dosimetry, with a performance comparable to that of commercially available detectors for SRT small field dosimetry, such as the EBT3 Gafchromic films and the PTW microDiamond.

The air gap used to render the Octa a ‘correction-free’ dosimeter for OFs measurements was found to be applicable to both flattened and FFF beams, in accordance to previous studies in the literature.

For any given field size, the Octa allows for the simultaneous real-time read-out of OF and dose profiles for cross-plane, in-plane and two diagonal directions.

PDD values for all beam qualities investigated were accurate within 2%. Though a DPP dependence was found that could be corrected for, the high doses per pulse typical of FFF beams were not detrimental to the overall performance of the dosimeter.

Our conclusion was that the Octa, thanks to its a sub-millimetre pitch and 4 intersecting linear arrays, while still offering a stable and real-time readout provides a much more detailed 2D dose map characterization than that of its predecessor the MP512 and the Duo. At the same time, the monolithic silicon array detector technology developed by the CMRP on which the Octa is based, was proved to have unique characteristics for relative dosimetry applications for a wide range of beam qualities and dose per pulses.

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