STEREOTACTIC RADIO-SURGERY DOSE QUALITY ASSURANCE: A COMMERCIAL PLASTIC SCINTILLATION VERSUS A DIAMOND DETECTOR

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♦ Purpose

To evaluate a commercial plastic scintillation detector (PSD) versus a commercial diamond detector (DD) for the dosimetric pre-treatment quality assurance (QA) of stereotactic radio-surgery cranial treatments delivered with Dynamic Conformal Arcs (DCA).

♦ Methods & Materials

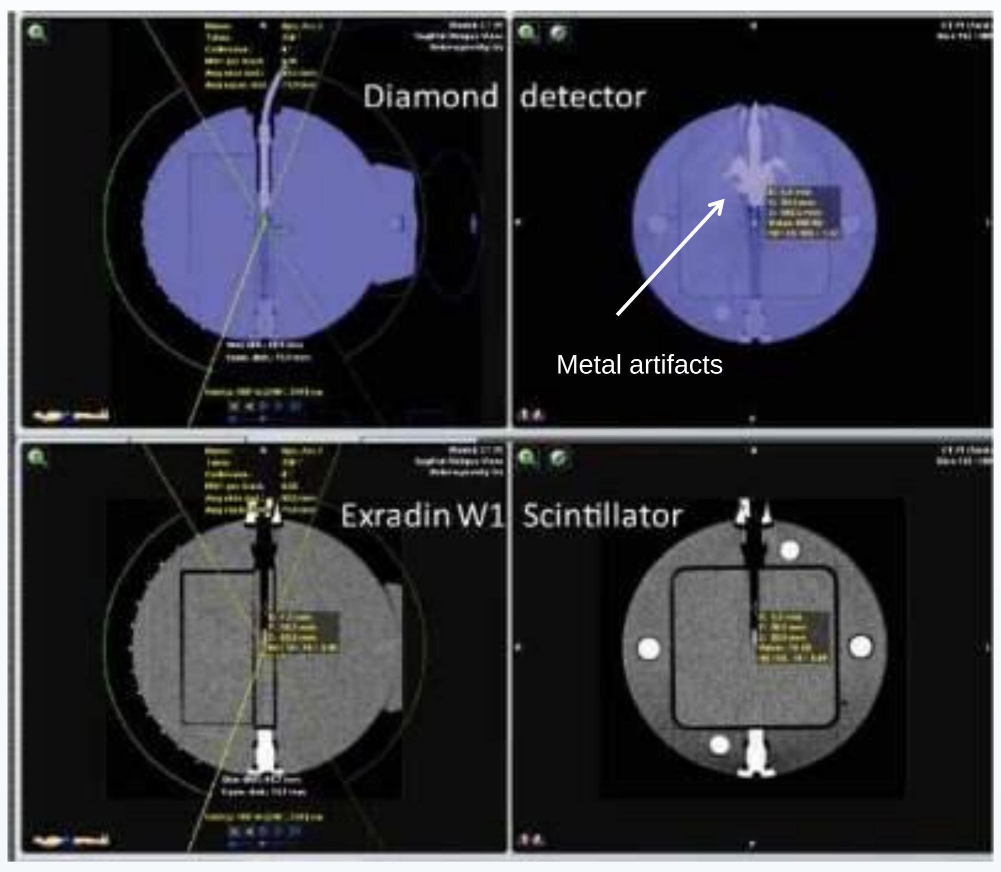

Figure 2. In blue the Lucy phantom with the DD and in grey, bottom, with the PSD, position "tip down".

A total of 15 patients plans and 79 DCA generated in iPlan RT Dose 4.1.2 (BrainLAB, AG, Germany), with a Pencil Beam algorithm using a kernel resolution of 2.5mm were tested. Mean equivalent MLC field size was of 16.6 mm (range 7.5-32.7mm). Plans were delivered on a Novalis TX, using the 6XSRS energy mode with a D/R of 1000 MU/min and a HDMLC multileaf collimator (2.5 mm leaf size at isocenter). Using a Lucy® 3D QA Phantom (Standard Imaging, Inc., USA), dose distributions were recalculated for each plan and the absolute isocenter dose was checked for each treatment field in the phantom, both with a DD (PTW 60003 Diamond Detector PTW, Freiburg, Germany) and with a PSD (Exradin W1 scintillator, Standard Imaging, Inc., USA), Figure 1. Table rotations, as present for the treatment, were implemented as part of our standard QA protocol. The detectors were positioned in the same way in the phantom, each with its specific insert, both with their longitudinal axis parallel the radiation beam when the gantry is at zero angle (see figure 2). This position (tip down Fig.2) was used to acquire the CT images of the phantom, define the isocenter location and the setup measurements. Figure 2 shows also that the DD creates artifacts in the CT scan caused by the metallic connectors of the detector which can lead to dose under-dosage estimation for table position 0° and arcs passing though the artifacts, that is gantry 0°. This effect is not present for the PSD. To overcome this problem for these irradiation configuration, plans were recalculated on phantom CT images with the detectors tip inserted in opposition respect to the beam entry direction (tip up). Measurements were also performed in this configuration, Fig.3.

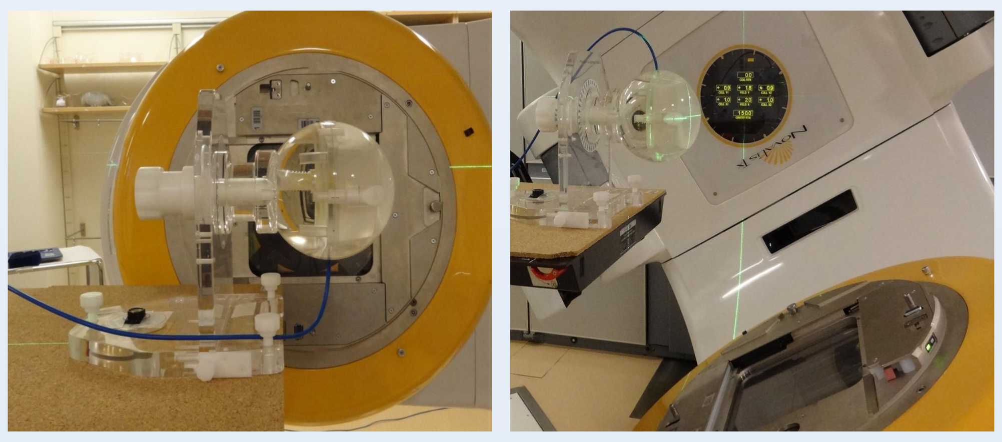

Figure 3. DD inserted in Lucy phantom for measurements: left tip up, right tip down.

Our DD, because of its small detector's volume, 0.0014 cc against 0.0024 cc of the PSD, is being used as our gold standard for fixed dose rates treatments. Furthermore our DD presents a well centered volume, as visible on X-ray images Fig.4 a-b. Figure 4 c, acquired in the same condition as Fig4 a-b, shows the absence of metal in the PSD. Figure 4 d shows images acquired in water of different detectors, including the W1 visible only at very low energies (30 kVp!).

	DD	PSD
TPS PBC algorithm	2.0%§	2.0%§
Absorbed dose determination uncertainty (Swiss protocol)	1.4%	1.4%
Daily linac output dose variation	"corrected"	"corrected"
Detectors accuracy	1.5%	1.0%*
Overall measurement accuracy (1SD)	**2.9%**	**2.6%**

Table 1. Overall measurement accuracy (1SD).

The overall accuracy of the measurements, 1 Standard Deviation (SD), is presented in Table 1.
Errors in repositioning the Lucy phantom, ≈0.05cm, are neglected in this study (error considered random because data was acquired on different days).

Calibrations to correct Cerenkov radiation† in the PSD, that use 40x40cm2 and 10x10 cm2 fields, were performed using the 6X 600D/R and not the 6XSRS, 1000 D/R, because SRS energy mode is limited to a maximum field sizes of 15X15 cm2.

We assumed that this calibration conditions was applicable, because the constructers reports an energy and dose rate independence response of the PSD. Calibration was made in solid water at a depth of 5cm with SSD set to 95cm.

Figure 1. W1 elements: (1) scintillator detector, (2) optic fiber, (3) Photo-detector, (4) 2 channel electrometer.

♦ Results

Mean % (measured/expected) ± SD		
	DD	PSD
All arcs	1.020 ± 0.035	1.020 ± 0.025
No table 0°arcs	1.014 ± 0.022	1.021 ± 0.026

Table 2. The mean dose ratios DD& PSD

The mean dose ratios (measured /expected) are presented in Table 2. Excluding DCA with table rotations at 0° (5 arcs) the mean dose ratio and SD changed for the DD because we excluded the false expected doses (caused by the DCA entering though the artifacts in the phantom).

The water equivalence and absence of metal in the PSD resulted in no change in data for this detector. Both DD and PSD results were within the overall measurements accuracy.
Analysis of the measurements performed with the detectors tip up and down, Fig.3, are shown below in Table 3.

		% Ratio Meas. down/up		% Meas./ exp. tip down		% Meas./ exp. tip up		Diff. down-up Meas./ exp.	
StartStop (arc - length)	Dose (Gy) per 1° arc	PSD	DD	PSD	DD	PSD	DD	PSD	DD
340° to 20° (40°)	0.106	2.0	0.5	4.7	24.1	4.8	0.2	-0.1	23.8
330° to 80° (120°)	0.046	1.3	2.0	1.1	8.9	0.4	-1.5	0.7	10.4
300° to 60° (120°)	0.052	1.4	1.0	1.8	7.8	0.7	-1.3	1.1	9.1
280° to 350° (70°)		-	-	1.5	1.6	-	-	-	-
280° to 70° (150°)	0.049	1.0	1.3	0.0	5.6	-0.6	-2.2	0.6	7.8

Table 3. Analysis for doses acquired at table 0°.

From the 3rd and 4th column it is possible to see that no dependence on detectors orientation was seen for both devices when measuring even with radiation coming mostly parallel through their connector side. Sixth and 8th columns show the effect of the DD metal artifacts in phantom and how they can affect estimated dose values. This effect is strongly depending on the amount of radiation passing though the artifacts (dose per angle°, around 0° gantry), 1st and 2nd column.

For the DCA with 70° length, since it is not passing thought the 0° gantry position, but remains at 10° from it, no dose perturbation was recorded and no tip up measurements was performed.

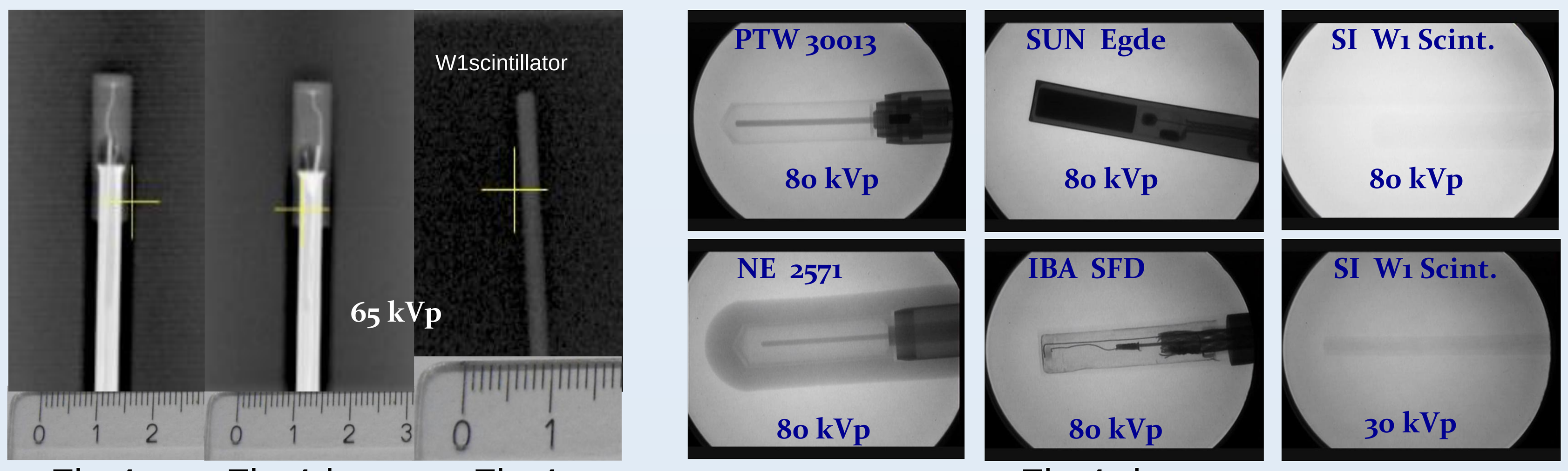

Fig 4-a. Fig 4-b. Fig 4-c. Fig 4-d.

Figures 4. (a-b) DD orthogonal views (65 kVp) in air, (c) W1 view (65 kVp) in air, (d) ionization chambers and W1 view (80kVp) in water. W1 is visible only with 30kVp X-ray.

♦ Conclusions

The Exradin W1 seems to be suitable for pre-treatment QA of typical clinical MLC based stereotactic radio-surgery plans. Furthermore, the Exradin W1 Scintillator because of its water-equivalence did not cause CT artifacts when scanned within the phantom, allowing any desirable configuration to be used.

♦ References

***§**Chow, J.C.L. et al. Med. Phys. 30, 2686 (2003); * Beddar, S. Private communication;*
***†** Guillot et al. Med. Phys. 38, 2140 (2011)*

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