Original paper

Dose calculation accuracy in proximity of a pacemaker: A multicenter study with threecommercial treatment planning systems

Anna Delana^{a,*}, Agnese Barbareschi^b, Rita Consorti^c, Maria Daniela Falco^d^a Department of Medical Physics, S. Chiara Hospital, APSS Trento, Italy^b Department of Medical Physics, ASST Spedali Civili di Brescia, Brescia, Italy^c Department of Medical Physics, San Filippo Neri Hospital ASL Roma 1, Rome, Italy^d Department of Radiation Oncology, "G. D'Annunzio" University, "SS. Annunziata" Hospital, Chieti, Italy

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ABSTRACT

This study compares Treatment Planning System (TPS) out of field dose calculation on a pacemaker (PMK) during external beam radiotherapy treatment. We consider four TPSs (Elekta-Monaco, Oncentra-Masterplan and two Philips-Pinnacle3) commissioned for two linacs (Elekta Sinergy and Varian Clinac) delivering two test beams (a highly modulated one and a square field) and two clinical breast plans. To calculate and measure dose to a PMK we built a Real Water3 phantom with a PMK embedded in it. Measures are performed with thermoluminescent dosimeters and Mosfet dosimeters. We evaluate differences between TPS calculated values for the dose to the PMK (both point dose and dose-volume histogram parameters) when the PMK is positioned in the first 10 cm outside the radiation fields. TPS calculation accuracy is evaluated comparing such values with measures. Differences in TPS calculations are on average 3.5 cGy Gy^{-1} for the modulated beam, and always lower than 2 cGy Gy^{-1} for the square beam. TPS dose calculation depends mostly on the TPS algorithm and model rather than the linac commissioned. TPSs considered show different degrees of calculation accuracy. In the first 4 cm to the field edge three out of four TPSs are in good agreement with measurements in the square beam, but only one keeps the agreement in the modulated beam: the others show over and underestimations up to $+20\%$ – -40% . The same accuracy is found considering a homogeneous phantom. Our results confirm what reported in previous studies and highlight the impact of TPS commissioning.

1. Introduction

The use of implantable cardiovascular electronic devices (CIED) such as pacemakers (PMK) and defibrillators (ICD) has increased over the last decades. An increasing number of patients with CIEDs is therefore referred to radiotherapy centers as both conditions, the “dependency on a CIED” and “cancer diagnosis”, are more frequent with older age and therefore overall frequency increases in an aging society [1].

When exposed to high energy radiation, CIEDs may encounter various malfunctions at doses exceeding 2 Gy, such as reset to back-up mode, deprogramming or signal interference even if the devices are not directly exposed to the radiation field [2,3]. Neutrons produced by the interaction of photon beams at energy $>10 \text{ MV}$ with heavy metals in the LINAC head are believed to be the main cause of CIED defects [3,4]; therefore, many guidelines consider a maximum energy of 6 MV optimal for treating patients with CIEDs [5–10].

Clinical guidelines [5,8–10] provide guidance for patient management and identify risk classes considering both clinical and dosimetric aspects of the radiotherapy (RT) treatment plan. Low risk is associated with patients that are not pacemaker-dependent and with CIED maximum doses $<2 \text{ Gy}$. Medium risk is assigned to patients not pacemaker-dependent with CIED maximum doses between 2 Gy and 10 Gy and patients pacemaker-dependent receiving a CIED maximum dose less than 2 Gy. Patients who receive a CIED maximum dose of 10 Gy or above are classified as high risk. Thus, the dose delivered to the CIED plays an important role in the categorization of these patients and medical physicists must carefully estimate it. The aforementioned guidelines suggest that, in order to diminish the dose received by the CIED, it should be kept outside the primary radiation field [2,5,8–10].

The estimation of CIED dose should be performed with the use of “in vivo” dosimetry and/or a treatment planning system (TPS) [7]. As reported in AAPM TG 158 [11], TPSs may show large calculation errors at

* Corresponding author.

E-mail address: anna.delana@apss.tn.it (A. Delana).<https://doi.org/10.1016/j.ejmp.2020.10.015>

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distances >3 cm from the field edge and/or below the 5% isodose line, i.e. in the region where a CIED is typically located relative to the target volume [6,7]. In the first 10 cm from the field edge the components of the out-of-field dose are, in order of relevance, patient scatter, collimator scatter and head leakage. Particularly the last two components are likely not being modelled properly by the TPS [12]. As stated in AAPM 203 [8] another source of error arising in the TPS's calculation is the presence of high-density objects and CIEDs usually have metal components and a metal case.

TPS dose calculation accuracy in the out-of-field region has already been investigated [12–16] but specific TPSs dose error estimation in proximity of a high-density object and in particular relative to CIED structures have not yet been performed [17]. As medical physicists usually consider dose volume histograms (DVH) to estimate the maximum and mean dose received by CIEDs, DVH dose calculation errors must be evaluated too. Most reports concerning out of-field-dose [12–16] consider only one TPS commissioned for one linac (mostly Varian Clinac), therefore there is a need for a study that compares several TPSs and considers Elekta linacs as well.

The aim of this multicenter study is to compare out-of-field dose calculation results in proximity of a CIED for a variety of TPSs currently in clinical use as well as the calculation of CIED DVHs (CIED maximum dose and mean dose). The second part of this work then evaluates CIED dose calculation accuracy when compared to TLD- and MOSFET-based phantom measurements.

2. Materials and methods

2.1. Phantom, imaging, and treatment planning

To reproduce the position of the CIED, in this case a pacemaker (PMK) in the human body and to measure the dose to the PMK, we used an in-house built phantom. The phantom consists of a PMK embedded in a PMMA slab with 2 cm of PTW Real Water 3 (RW3) positioned above (CIEDs are usually placed between 1 and 3 cm beneath the skin [8]) and 10 cm of RW3 below (Fig. 1). The PMK was covered with wax to avoid air gaps and to host three holes for the dosimeters to accurately reproduce their positioning during dose verification at the linac (Fig. 1 (a)). These holes were placed above the PMK: one approximately at its proximal end (P1), one in the middle (P2), one at the distal end (P3). CT images of this phantom were acquired with a Philips Brilliance BigBore (120 kVp, 250 mAs/slice, 3 mm slice thickness). Center C1 delineated the structures PMK, CT artifacts and the positions P1, P2 and P3. CT datasets and structures were then sent to all participating centers in DICOM format. Density correction was performed only for CT artifacts. For the PMK each TPS used its CT to ED calibration curve and calculation grid used in clinical practice (for details see Table 1).

Three commercial TPSs commissioned for two linac models were included in the study. Table 1 summarizes TPS, dose calculation algorithms, maximum value of CT-ED calibration curve, calculation grid resolution used and linacs for which they were commissioned. TPS P3E is currently used only for 3D conformal planning, while the other TPSs are also used for Intensity Modulated RT (IMRT) and volumetric modulated arc therapy (VMAT) plans.

2.2. Test plans

To compare out of field dose values calculated by different TPSs for non-modulated and highly-modulated beams, we considered two simple test plans that could be easily created on each TPS. The non-modulated beam was a 10×10 cm² square beam. The highly-modulated beam was a 10×10 cm² modulated field as a series of 10 rectangular strips in the linac's AB direction of 10 cm $\times$ 1 cm each and a beam modulation (BM) of 0.9 as defined by Du et al [18]. Each plan had a prescription dose of 2 Gy at the isocenter, at SSD 95 cm and 6 MV energy. Dose calculation is performed by shifting the beam isocenter in the linac's TG direction, in

order to have the PMK proximal edge from 1 cm to 10 cm from the field edge (see Fig. 1). Center C1 also analyzed a clinical case of a left side breast cancer patient simulating two more plans: a 3D conformal plan with field in field technique, BM of 0.29 (with TPS MC_E and P3_E) and an IMRT plan with two dMLC beams and BM of 0.5 (with TPS MC_E). The plans were calculated and optimized on a patient CT dataset, prescribing 50 Gy in 25 fractions, and then transferred and recalculated on the PMK phantom with the beams collapsed at gantry fixed at 0°.

2.3. Phantom irradiation

Dose measurements were performed with Lithium fluoride thermoluminescent dosimeters (Harshaw TLD-100 LiF, square rods with dimension of $6 \times 1 \times 1$ mm³) at center C1 and Thomson Nielsen's mobile MOSFET Dose Verification System (TN-RD-70-W) with MOSFET dosimeters (TN-502RD, nominal sensitivity 1 mV/cGy) at center C3. These detectors have been shown to be accurate as out-of-field dosimeters operated at 6MV [20–22]. Detectors were inserted at each hole P1, P2 and P3, and measurements were repeated shifting the phantom in the linac's TG direction to measure the out of field dose at increasing distance from the field edge (1 cm, 3 cm, 5 cm and 10 cm).

2.3.1. Detector calibration

For each measuring session, detector calibration was performed in RW3 slabs with a 10×10 cm² square 6 MV beam, prescribing 2 Gy at the isocenter.

Before each calibration, TLDs were properly prepared and background radiation was estimated with 3 unexposed dosimeters and subtracted from the exposed TLD reading. TLD batch homogeneity response was less than 3%. TL detectors are energy-dependent [23,24] and additional correction factors should be used to reduce the over-response at low energy spectrum. However, at the distances and depth considered in this experimental setup, TLD sensitivity was previously estimated to be close to unity [25], therefore we did not apply further corrections.

MOSFET calibration was performed with the 3 detectors used and each measurement was repeated 5 times. Before each measurement session, background radiation was estimated and taken into account by the measuring system. MOSFET reproducibility depends on dose levels and their age, reproducibility at dose levels of interest for this study ranges from 4% (at 10 cGy) to 15% (in the range 2–4 cGy) [21].

2.3.2. Detector uncertainty estimation

To achieve the highest possible accuracy for the measured dose, the signal was quantified as the average of repeated measurements (5 TLDs at each hole, MOSFET measures are repeated 3 times) and the uncertainty of the measurements was estimated as the standard deviation of the readings. The error due to the uncertainty of the positioning of the dosimeters was quantified with the formula:

$$error\% = dosegradient \left[\frac{\%}{mm} \right] * positioningerror[mm] \quad (1)$$

where the dose gradient is estimated as the maximum difference between $D(x)/D_{CAX}$ measured in the first 3 cm from the beam edge and the positioning error is 3 mm. The resulting dose uncertainty due to positioning error was less than 1%. We also estimated the accuracy of the dosimeters at low doses by analyzing their reproducibility in the dose range of interest in our study.

Overall uncertainty, considering setup errors and low dose measure repeatability, was quantified as 9% for the TLD and ranging between 6% and 17% for doses between 2 and 12 cGy for MOSFET.

3. Analyses performed

All analyses were performed in terms of $D/D_{CAX}(x)$, i.e. the dose at distance x to the beam edge normalized to the central axis dose $D(CAX)$

Fig. 1. (a) Schematic drawing of the phantom with the PMK. (b) PMK embedded in wax; (c) and (d) CT image of the phantom with the structures delineated (coronal and sagittal views). In red the PMK structure, in gray the CT artifact and in blue, green, and yellow the structures of dosimeters positions P1 (yellow), P2 (green) and P3 (blue). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

and expressed as % of D_{CAX} .

3.1. TPS dose comparison

3.1.1. PMK dose analysis

TPS dose comparison was performed in terms of dose to the PMK. We considered dose to P1, P2 and P3 as reported in Fig. 1 and, for PMK's DVH analysis, dose to 1 cc of PMK structure (D_{1cc} i.e. the maximum dose received by the PMK), and mean dose to the PMK (D_{mean}). For each PMK position x and each parameter considered (point dose or DVH parameter) we considered the maximum difference $\Delta D_{MAX}(x)$ between TPSs' calculated dose values

$$\Delta D_{MAX}(x) = \max_{i,j \in \{1a,1b,2,3\}} |(D/D_{CAX}(x))_{Ti} - (D/D_{CAX}(x))_{Tj}| \quad (2)$$

where the indexes i and j run over all TPSs combinations. These analyses were also performed by separating the analysis of the maximum absolute difference between $D/D_{CAX}(x)$ calculated from TPSs with a CCC calculation algorithm for the Elekta linac ($|P3_E-MP_E|$), and the two Philips – Pinnacle3 TPSs commissioned for different linacs ($|P3_E-P3_V|$). Analyses done for the test plans are performed also for the 3D-CRT clinical breast beams considering TPSs MC_E and $P3_E$.

3.1.2. Comparison with homogeneous phantom

We evaluated the impact of the PMK's electron density on dose calculation calculating D/D_{CAX} with each TPS on a homogeneous RW3 phantom at the same distances to the field edge. Then, we analyzed the difference in the D/D_{CAX} values estimated from each TPS on the two phantoms. As both $P3_E$ and $P3_V$ are Pinnacle3 TPSs, only $P3_V$ was considered for this analysis.

3.2. TPS dose accuracy

3.2.1. PMK dose analysis

TPS accuracy was quantified as the ratio of calculated to measured values $\frac{D_{TPS}}{D_{meas}}$

$$\frac{D_{TPS}}{D_{meas}} = \frac{(D/D_{CAX})_{TPS}}{(D/D_{CAX})_{measure}} \quad (3)$$

To estimate DVH parameter calculation accuracy, we first estimated the Person's correlation coefficients between D_{1cc} and the calculated P1 dose and between D_{mean} and the mean of the calculated point doses. We evaluated TPSs DVH accuracy comparing measured dose at P1 with D_{1cc} .

We also investigated if beam modulation affect TPS calculation accuracy.

3.2.2. Comparison with homogeneous phantom

We compared TPS out-of-field dose accuracy in presence of a PMK and in homogenous medium by quantifying $\frac{D_{TPS}}{D_{meas}}$ for point dose in the two phantoms, for each TPS and for the square and IMRT beams.

4. Results

4.1. TPS dose calculation comparison

4.1.1. PMK dose analysis

Fig. 2(a) shows the point dose calculated by the TPSs and Fig. 2(b) shows ΔD_{MAX} , $|P3_E-MP_E|$ and $|P3_E-P3_V|$ as a function of the distance from the beam edge for the IMRT field. Discrepancies between all TPSs (ΔD_{MAX}) have a mean value of 3.5% (+/- 0.7%, 1SD) and are higher far

Table 1

TPSs and linacs considered in our study. No relevant changes in calculation algorithm nor in LINAC modelling had been implemented between Pinnacle3 versions considered. XVMC: Voxel Monte Carlo [19]; CCC: Collapsed Cone Convolution.

Centers	TPS		Algorithm	Max CT to ED value	Calculation Grid	LINAC
C1	MC _E	Elekta Monaco v.5.11	XVMC	3.8	2 mm	Elekta Synergy Agility
C1	P3 _E	Philips Pinnacle3 v.9.8	CCC	4.6	2 mm	Elekta Synergy Agility
C2	MP _E	Elekta Oncentra Masterplan v.4.5.3	CCC	2.8	2 mm	Elekta Synergy Agility
C3	P3 _V	Philips Pinnacle3 v.9.10	CCC	1.8**	3 mm**	Varian Trilogy

** TPS P3V performed the calculations also with a 2 mm grid and CT to EC density curve of 4.6 like the other Philips Pinnacle TPS P3_E.

from the beam edge; $|P3_E - MP_E|$ and $|P3_E - P3_V|$ are always less than 2% and $|P3_E - P3_V|$ is always lower than 0.5%. Higher ΔD_{MAX} values are related to TPS MC_E (XVMC algorithm) with respect to the CCC TPSs: XVMC calculates a higher dose with respect to the other TPSs.

ΔD_{MAX} values for the square field are always less than 2% (not reported in a figure) and are higher closer to the beam axis. At distances greater than 4 cm ΔD_{MAX} decreases to an almost constant value of 0.5%.

In Fig. 3 PMK D_{1cc} are reported as a function of the PMK's distance from the beam edge; PMK D_{1cc} are normalized to the central axis dose and differences between TPSs values are not shown. In the square beam, D_{1cc} shows almost the same behavior for all TPSs. D/D_{CAX} differences are lower than 1.5% when considering all TPSs and lower than 0.5% when considering Pinnacle3 TPSs. Comparing TPSs commissioned to the same linac model in the IMRT field, MC_E shows the same behavior as TPS P3_E up to 7 cm with a maximum difference of 0.7%, and it calculates a higher dose than MP_E.

D_{mean} has a similar behavior to D_{1cc} (graphs are not reported) but differences between TPSs are lower: for the square beam TPSs have discrepancies of less than 1% of D_{CAX}, and less than 3.5% of D_{CAX} in the IMRT field. D_{1cc} for the two breast plans has the same behavior as the square test beam. Differences in the TPSs' calculated D_{1cc} parameter are greater closer to the field edge, with a maximum value of 1.7%, while differences in the calculation of D_{mean} are almost constant with a value of 0.2%.

4.1.2. Comparison with homogeneous phantom

Fig. 4 shows the out-of-field D/D_{CAX} calculated in the two phantoms as a function of the beam edge distance. Differences in the D/D_{CAX} values calculated on the two phantoms are almost negligible for both

fields and always less than 1.6%.

4.2. TPS dose calculation accuracy

4.2.1. PMK dose analysis

We compared TPS- calculated D/D_{CAX} for positions P1, P2 and P3 with measurements as shown in Table 2. TPS MC_E is in very good agreement with measurements for the square beam, whereas for the IMRT plan the agreement is worse. TPS P3_V has good agreement between measured and calculated D/D_{CAX} for both plans.

For the two breast plans (3D-CRT and dMLC) TPS MC_E calculates a slightly higher D/D_{CAX} than the measured value, while for TPS P3_E this is true only for D/D_{CAX} greater than 1.5%.

Fig. 5 shows the average value of $\frac{D_{TPS}}{D_{meas}}$ calculated over all the POIs considered as a function of BM. To further compare our results with the study of Huang et al. [12], we plot them as a function of MU/Gy.

4.2.2. DVH dose analysis

Pearson's correlation coefficient between P1 dose and D_{1cc} ranges from 0.97 to 0.99 so we used P1 dose measurements to evaluate the accuracy of this DVH parameter. Pearson's correlation coefficient between the average value of P1, P2 and P3 doses and D_{mean} ranges from 0.83 to 0.99, therefore they are not linearly correlated, and we evaluated calculation accuracy only for the calculated point dose and not for this DVH parameter.

Table 2 also shows the average ratio of $\frac{D_{TPS}}{D_{meas}}$ values for P1 that we took as the reference dose for D_{1cc}. Both fields have the largest discrepancy far from the field, with the PMK at 5 cm to the beam edge.

Fig. 2. (a) Point dose at different distances from the beam edge for the IMRT beam (b) Maximum difference between calculated isodose at all points for the IMRT plan. Blue dots represent the maximum discrepancy between calculated isodose of all TPSs (ΔD_{MAX}), light blue dots are the difference between TPSs with two CCC calculation algorithms commissioned for the Elekta linac ($|P3_E - MP_E|$), and red dots are the difference between Pinnacle3 TPSs commissioned for two different linacs ($|P3_E - P3_V|$). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Fig. 3. D_{1cc} at various PMK distances from the field edge, as calculated by TPSs included in our study for the square (a) and the IMRT (b) 10x10 cm² beam. Values of D_{mean} have the same behavior.

Fig. 4. Calculated values for out-of-field D/D_{CAX} [% of D_{CAX}] for the square field (a),(c) and for the IMRT field (b),(d) in the RW3 homogeneous phantom (blue lines) and in the heterogeneous phantom, with the PMK (red lines). Diamonds represent measures with error bars. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

4.2.3. Comparison with homogeneous phantom

Fig. 6 shows the average ratio of $\frac{D_{TPS}}{D_{meas}}$ on the two phantoms considered (homogeneous RW3 and with the PMK) for TPSs MC_E , MP_E and $P3_V$. In the PMK phantom only P1 is considered. Values on the two phantoms are within 1SD from each other (except MP_E in the square field, 2SD) for both fields.

5. Discussion

Our study aimed to compare TPS dose calculation in the evaluation of dose to a PMK, considering both out of field dose points and DVH dose estimations. We investigated their accuracy by comparing calculated doses with measurements. We also evaluated the impact of the introduction of a high-density object such like a PMK on the accuracy of TPS calculation.

Table 2

Average $\frac{D_{TPS}}{D_{meas}}$ for all beams and plans considered. Measuring errors (in brackets) at corresponding dose levels are reported. Bold numbers highlight a good agreement between calculated and measured values.

	Square beam			IMRT beam		
	1–4 cm	4–10 cm	P1	1–4 cm	4–10 cm	P1
MC_E	1.05 (±9%)	0.99 (±9%)	0.99 (±9%)	1.18 (±9%)	1.53 (±9%)	1.23 (±9%)
P3_E	0.97 (±9%)	0.65 (±9%)	0.84 (±9%)	0.62 (±9%)	0.54 (±9%)	0.59 (±9%)
MP_E	0.64 (±9%)	0.69 (±9%)	0.72 (±9%)	0.65 (±9%)	0.83 (±9%)	0.67 (±9%)
P3_V	1.01 (±15%)	1.18 (±15%)	1.07 (±17%)	0.97 (±12%)	0.86 (±13%)	0.93 (±13%)
3D-CRT breast plan				dMLC breast plan		
MC_E	1.10 (±9%)	0.99 (±9%)	1.05 (±9%)	1.09 (±9%)	0.85 (±9%)	0.95 (±9%)
P3_E	1.12 (±9%)	0.76 (±9%)	0.93 (±9%)	–	–	–

In the calculation of the out-of-field dose, discrepancies between TPS's dose calculation are larger for the IMRT field than for the square field: the maximum difference found for the IMRT field is 4.7% whereas for the square field, differences are always lower than 1.5%. When considering only CCC TPSs, differences are always lower than 2% regardless of the field considered. As shown in Fig. 2a, each TPS model has its own behavior. The two Pinnacle3 TPSs (P3_E and P3_V) show a steep exponential decrease with distance to the CAX, MC_E shows a slower decrease in dose with distance and MP_E returns a constant value of 2.7% after 4 cm to the field edge. Such different behaviors are caused by differences in the calculation of the patient and collimator scatter components by the TPSs. As reported previously [12,26,27], type B TPSs underestimate these dose components when compared with Monte Carlo simulation. Collimator scatter strongly depends on the modulation of the field. The modulation of the IMRT field in our study is about 10 times greater than in the square field and therefore the biggest differences are found in the dose calculated for the IMRT field.

TPSs dose calculation accuracy was estimated by comparing dose calculated over the PMK structure with measurements, increasing the PMK's distance from the beam edge (results are reported in Table 2). TPSs MC_E and P3_E show a better agreement with measurements in the square field and in the breast plans than in the IMRT field. TPS MP_E always underestimates the measured dose and TPS P3_V is always in good agreement with measurements. In the first 4 cm from the beam edge of the square beam (where the doses received by the PMK are of clinical relevance) all TPSs except MP_E are in good agreement with measured data. At distances greater than 4 cm from the beam edge only TPSs MC_E and P3_V dose calculations are within our measurement error for the low-modulated beams (square beam and breast plan). TPSs with CCC (apart from P3_V) mostly underestimate the measurements, while the TPS with the XVMC dose calculation algorithm (MC_E) mostly overestimates measurements. Deviations between calculated and measured dose values are almost always within acceptability criteria reported in IAEA TRS430 for the out-of-field dose region [28,29]. Our results with CCC TPSs are in accordance with previous studies that reported a TPS's

underestimation in the out-of-field dose region by up to 50% [11–13,15,16,30]. No studies have been performed with the XVMC calculation algorithm yet. Joosten compared a Monte Carlo model (BEAMnrc) with TLD measurements, and found differences in the range of [–3%, +23%] in the first 10 cm from the field's edge, confirming our results of TPS's dose overestimation [27].

The two Pinnacle3 TPSs show different behaviors: P3_V is always in good agreement with measured data, while P3_E always underestimates the measurements. We investigated if this difference in TPS accuracy was related to calculation grid size or maximum CT-ED value. TPS P3_V recalculated dose distributions with calculation 2 mm grid size and with maximum CT-ED value equal to 4.6, as TPS P3_E: POI doses change at maximum of 0.04% of D_{CAX} and DVH parameters D_{1cc} and D_{mean} differ at maximum of 0.3% of D_{CAX}. Therefore the impact of the grid size and maximum CT-ED value on the TPS accuracy estimation is less than 1%.

Fig. 6. Average $\frac{D_{TPS}}{D_{meas}}$ values for the two test fields and the two phantoms. Error bars represent 1SD of the data.

Fig. 5. TPS calculation accuracy as a function of beam modulation (a) and MU/Gy (b). Error bars represent 1SD of the data.

Another aspect that we analyzed, is the accuracy of linac modeling for the two TPSs, during commissioning process: we quantified it by calculating the mean error between the measurements and calculated out-of-field doses. This error is 10 times lower for center C3 when compared to C1. The better performance of TPS P3_v could be related to a finer tuning of this TPS in the out-of-field region, during commissioning.

The fact that the TPS MC_E - Monaco with XVMC calculation algorithm - shows a worse accuracy in the out-of-field dose calculation than TPS P3_v - Pinnacle3 with CCC calculation algorithm- can be explained by the fact that Pinnacle3 TPSs are tunable by the user, while this is not possible in Monaco. However, MC_E dose calculation is a conservative estimate of the dose and such discrepancies with measured dose would not lead to clinically relevant issues. TPS MP_E is not tunable by the user either, and shows an underestimation of the measured dose of an average 35%.

Huang [12] in his study also considered plan modulation as a possible source of TPS's inaccuracy in out-of-field calculation but found no difference between plans with different modulation. Our results partially disagree with his results, as we find poorer calculation accuracy in the highly modulated beam and a good agreement with measurements for the low modulated fields for TPS MC_E (XVMC calculation algorithm). TPSs with a CCC algorithm are in accordance with Huang's results (as reported in Fig. 5b). We can therefore suggest that a dependency of TPS accuracy on plan modulation may be related to the calculation algorithm.

Following AAPM 203 guidelines [8], medical physicists usually consider DVH parameters to estimate the dose received by the CIED. We therefore studied how different TPSs' calculation affects DVH dose estimation and the accuracy of DVH dose parameters D_{1cc} and D_{mean}. We first evaluated the clinical impact of differences in the calculated D_{1cc} and D_{mean} (reported in Fig. 3) for a 25-fraction treatment (2 Gy/fraction at the isocenter). For the square field D_{1cc} and D_{mean} differences are on average 48 cGy and 34 cGy, while for the IMRT field these differences can exceed 200 cGy at all distances to the beam edge. Even between TPSs with the CCC algorithm commissioned for the same linac model (P3_E and MP_E), differences are on average 150 cGy for the IMRT field and 22 cGy for the square field. This variability in DVH dose estimations could impact the patients risk class definition and therefore individual patients' management, especially for highly modulated plans.

We also studied the accuracy of DVH parameter D_{1cc} by analyzing the accuracy of POI P1 dose. When the PMK is positioned at 1 cm to the beam edge the calculated P1 dose values are consistent with measured dose for all TPSs except MP_E, for the square plan, and only for MC_E and P3_v for the IMRT plan. The farther the PMK is positioned from the field edge, the worse agreement is found between D_{TPS} and D_{MEAS}. As shown in table 2, TPS P3_v is always within the dosimeter's uncertainties while TPS MC_E is in good agreement with measured values for the low modulated beams and overestimates the measurements by an average value of 20% for the IMRT beam. This means that D_{1cc} is accurately estimated by TPSs P3_v and conservatively estimated by TPS MC_E. Other TPSs underestimate the dose with respect to the measurements with an average value of 20% for the square field and 40% for IMRT. These underestimations, even if within tolerances of IAEA TRS430 for the out-of-field dose region [28], must be considered when D_{1cc} calculation is used for patient risk class assessment.

Tests performed in this study focused on two simple beams, which represent two extreme cases of plan's modulation and one clinical case (left breast treatment). As the highest source of uncertainty in the out-of-field region arises from head scatter modeling [12,27], our IMRT beam (modulation of 0.9) might represent an extreme situation with respect to most clinical cases. Moreover, we decided to perform our tests at 2 cm depth, as to be within typical CIED positions inside the body and to mitigate the dosimetric effect of positioning errors that could arise in the build-up region. We therefore may assume that dosimetric uncertainties of PMK dose in clinical plans should stay within our results.

By comparing TPSs' calculation in presence of a PMK and in

homogeneous medium, differences in the calculated D/D_{CAX} are almost negligible for all TPSs: for both fields the difference is always less than 1.6% of D_{CAX}, even though TPSs have different CT-ED calibration curves. For all TPSs considered and for both test beams, the agreement between the measured and calculated dose on the PMK phantom does not differ from that on the homogeneous RW3 phantom. $\frac{D_{TPS}}{D_{MEAS}}$ estimated on the RW3 homogeneous phantom are within 1SD from that estimated on the heterogeneous phantom, except for MP_E for which the agreement is within 2SD (see Fig. 6). We can therefore assume that the presence of a high-density object in the out of field region does not modify TPS calculation nor, as a consequence, TPS calculation accuracy.

6. Conclusions

We found that TPS dose calculation in the out-of-field dose region depends mostly on the TPS algorithm and TPS model rather than on the commissioned linac. TPSs with CCC calculation algorithm commissioned for different linacs are more likely to return the same dose values than TPSs with different calculation algorithms commissioned for the same linac, especially in highly modulated beams.

Regarding TPS calculation accuracy over the PMK, our results confirm what is reported in AAPM TG 158 and in previous studies [11–13,15,30,31] and show the impact of TPS commissioning: the TPS with the highest calculation accuracy in the out-of-field dose calculation is the one for which the finest tuning in the commissioning process had been performed. Moreover, TPSs' accuracy may depend on beam modulation and on PMK position with respect to the beam edge. These aspects affect the accuracy of DVH dose estimations, too, which are used in clinical practice to quantify the dose received by CIED and to define patient risk class.

Our suggestion is therefore to also perform fine tuning in the out-of-field dose region during TPS commissioning to reduce dosimetric uncertainties in the estimation of PMK dose or, if not possible, to estimate these uncertainties with in-vivo/in-phantom measurements. As not all calculation algorithms have the same dependency on beam modulation, a correct estimation of this dependency should be carried out, considering also highly modulated beams. When in-vivo measurements are not feasible, our study shows that measurements on a homogeneous RW3 phantom are suitable as the calculation accuracy in the out-of-field region is not affected by the presence of high-density objects.

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