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Commissioning of 6 and 10 MV Beams for Total Body Irradiation (TBI)

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ABSTRACT

The objective of the TBI treatment is the ablation of the bone marrow and the destruction of the circulating leukemia cells, once they are widely distributed throughout the body. Using beam parameters acquired under conventional SSD in TBI treatments may add non-negligible uncertainties in the monitor units calculation or in the beam profiles. The study, aims to commission the 6 and 10 MV photon beams, of the Varian accelerator CX model. A slab phantom and dosimetric assembly were used under TBI conditions. The accuracy of the TPS was evaluated against the experimental data. A set of data were acquired, highlighting the TPR table and methodology for calculating MU has been implemented. The TPS has presented a statistical uncertainty of ± 2.7 % compared to the experimental data for monitor unit calculation. The use of an acrylic spoiler has been shown to be clinically advantageous where, for a **6 MV beam, the entrance PDD was 75 % without a spoiler and 99.5 % with a spoiler. For a 10 MV beam, it was verified that without a spoiler, the entrance PDD was about 55 %, but with a spoiler, it was about 93 %. For medium Heterogeneous the TPS underestimated dose values by up to - 3.5 % with a mean deviation of – 2.9 %, for 6 MV and for 10 MV, the TPS overestimated the dose values by up to 1.1 %, with an average deviation of 1.0 % using the acrylic thorax phantom. The data obtained can be used clinically.**

Keywords: **TBI, Commissioning, Treatment.**

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1. INTRODUCTION

Total body irradiation (TBI) has been extensively used as a myeloablative and immunoablative method before allogeneic hematopoietic stem cell transplantation (HSCT) in patients diagnosed with acute lymphoblastic leukemia (ALL) for several decades. Fractionated TBI is commonly administered in most centers to mitigate acute side effects like nausea and vomiting, as well as late effects such as cataract. Moreover, a lung shielding is widely implemented to avert severe non-infectious pneumonitis [1].

Typically, myeloablative TBI is carried out in 8 to 12 fractions over a period of 4 days, with 2 to 3 daily treatments, and the most used dosage ranges from 12 to 15 Gy [2]. Administering doses above 15 Gy is recommended to reduce the transplant rejection rate. However, it is also associated with a higher likelihood of host disease and decreased survival rates over a 2-year period [3]. In TBI treatments, the dose rate plays a crucial role in the delivery of radiation. According to the American Association of Physicists in Medicine (AAPM) report TG-17, the recommended dose rate should be in the range of 6 to 15 cGy/min at midline of the patient. It has been observed that utilizing dose rates below 20 cGy/min is linked with a decrease in the incidence of complications [4].

There are multiple techniques utilized for delivering TBI, and the selection of technique depends on factors such as available equipment, photon beam energy, maximum field size, treatment distance, dose rate, patient size, and shielding requirements. The antero posterior technique (AP/PA) generally provides a more homogeneous dose distribution along the longitudinal axis of the body. Another frequently used technique is bilateral, where the patient is positioned on their back and exposed to lateral radiation beams. This approach provides greater comfort and, consequently, better reproducibility. However, it results in a greater variation in body thickness along the path of the radiation beam, leading to increased heterogeneity of the absorbed dose.

Compensators are necessary to achieve dose uniformity along the body axis within a tolerance of ± l0 % , although some non-critical structures and extremities may exceed this specification [5].

For radiation therapy treatments to be viable, it is crucial to characterize the radiation beam under various configurations that are clinically representative. Data commissioned in conventional SSD have limited use in TBI conditions as observed by various authors [6, 7], because do not

fundamentally contemplate the same pattern of scattering and backscattering because of inadequate size of the phantom, the field size and the set up measurement are totally different. The same can be said about the cable effect within the radiating field, and consequently the low-rate leakage current effect that is characteristic of TBI. Such limitations justify the characterization of the beam under TBI conditions to make the dose or monitor unit calculation more accurate.

The goal of this study is to commission the 6 MV and 10 MV photon beams for TBI and secundarily, evaluate the TPS accuracy against experimental data at the radiotherapy department of Brazilian National Cancer Institute (INCA)

2. MATERIALS AND METHODS

This research was conducted at the National Cancer Institute, where measurements were taken from Varian accelerator CX model using 6 MV and 10 MV external photon beams. The benchmark configuration for the study was established as having a source-surface distance (SSD) of 330 cm, a depth of 10 cm, a field size of 40 x 40 cm^2, the gantry at 90°, and the collimator set to 45° (figure 1). As observed in the figure 1, the IBA detector FC65P farmer with 0.65cc of volume calibrated in a secondary laboratory was placed in a solid water phantom with 30 x 30 cm^2 slabs, with a variable thickness. All measurement with ionization chamber (IC) was made it with this detector.

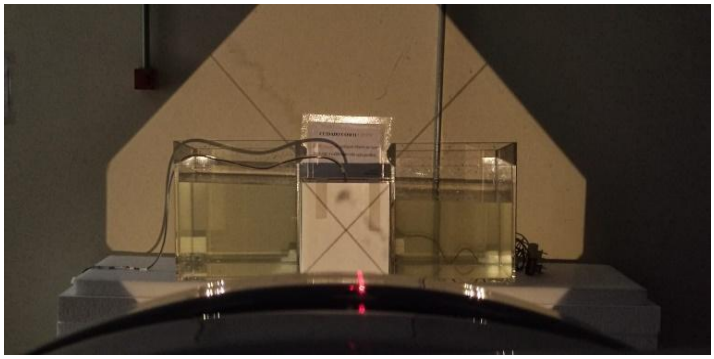
To replicate the scattering conditions found in treatments, additional phantoms were positioned laterally to produce similar conditions. The data was collected with and without the presence of a spoiler, an acrylic spreader sheet, with dimensions of 172 cm x 72 cm and 1 cm thickness. Its function is to superficialize the dose. While conventional radiotherapy aims to preserve the skin, in TBI it is preferred to have the entrance dose close to the prescribed dose. This is especially crucial for certain diseases like leukemia, where the cells can travel through the blood vessels in the skin [8].

The Brilliance CT Big Bore Radiology Scanner (*Philips*) was employed to scan the phantoms used in the study with a slice thickness of 3 mm. The *Gafchromic* EBT3 film analysis was done using the software Image J version 1.5

2.1 Experimental Set Up

The configuration shown in figure 1 presents the TPR (Tissue Phantom Ratio), dose profile, percentage depth dose (PDD), distance factor, and calibration measurements. Looking at the figure, it is possible to identify the solid water phantom flanked by water-filled acrylic phantoms.

Figure 1. *The configuration shown in figure 1 was used for measuring TPR (Tissue Phantom Ratio), dose profile, percentage depth dose (PDD), distance factor, and dose calibration."*



The TPR curve was obtained by varying the phantom thickness using solid water slabs and adopting 10 cm as the reference depth.

Dose profiles were obtained at depths of 5 cm and 10 cm. Due to the difficulty in performing this scan using conventional methodology, measurements were collected by moving the entire experimental apparatus laterally. The evaluation of symmetry and flatness was done at a depth of 10 cm. The PDD curve was obtained not only with IC, but also with gafchromic EBT 3 films. Measurements were performed with and without the presence of the spoiler, in order to assess its influence to the surface dose.

Additionally, the validity of using the inverse square law (ISL) was evaluated with reference configuration (figure 1) by varying the SSD to different values. The ratio of the doses was compared with the ratio of the inverse squared distances, as shown in equation 1.

$$\frac{\varphi_A}{\varphi_B} = \left(\frac{f_b}{f_a}\right)^2 \quad (1)$$

Where φ_A and φ_B are the photon fluences at distances f_a and f_b, respectively from S source.

Using ISL or not, the main parameters involved in the formalism of the MU calculation obtained in standard radiotherapy can be alternatively generated by conversion through their mathematical relationships, as indicated in equation 2 for example, a methodology that was adopted by several authors [5,6] when dealing with TBI commissioning. But a validation of the data under treatment conditions is indispensable.

$$TPR(d,r_d) = \frac{PDD(d,r,SSD)}{100\%} \times \left(\frac{SSD+d}{SSD+d_0}\right)^2 \times \frac{S_P(r_{d_0})}{S_P r_d} \quad (2)$$

Where, $\frac{S_P(r_{d_0})}{S_P r_d}$ is the phantom scatter factor ratio, d and d_0 are, the reference depth and the máximum depth dose, respectively.

Dosimetry in the TBI scenario includes a length of cable connected to the detector inside the radiation field. For the purpose to investigate, slabs phantom was placed on accelerator table with 100 cm of SSD, field size of 20 x 20 cm^2 and the influence of the cable within the radiation field was assessed by the ratio of readings with and without the presence of the cable.

Typically, the dose rates used for TBI treatments are the lowest available. In order to evaluate the dose rate constant, measurements were made at the following dose rates 100, 200, 300 and 600 MU/min under TBI configuration (figure 1). The reference dose rate was 100 MU/min.

In order to determine the calibration factor (cGy/MU) under TBI conditions, absolute dose measurements were taken at the defined reference conditions (SSD = 330 cm, field 40 x 40 cm^2, depth 10 cm and was delivered a 100 monitor units), figure 1. The calibration factors were obtained with and without the presence of a spoiler, for 6 and 10 MV beams.

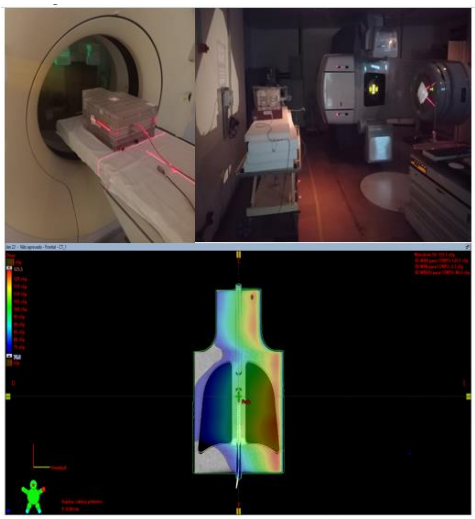
The output factors (OF) of fields 5 x 5 cm^2, 20 x 20 cm^2 and 40 x 40 cm^2 were measured using the reference set up as showed in the figure 1. The field sizes were defined at isocenter, SSD 100 cm.

2.2 Treatment Planning System (TPS) Perfomance check

The calculations algorithms are indispensable tools in dose calculation. Their precision is not only associated with the mathematical method, which can be convolution, convolution / superposition, or even Monte Carlo, but also with the modeling process. Given the complexity of the calculation, validation is required in different clinical settings ranging from the simplest to the

most complex. In the present study, the analytical anisotropic algorithm (AAA) based on the convolution-superposition mathematical method installed in Eclipse (Varian Medical Systems, Palo Alto, CA) version 13.6 was used [9]. The accuracy of AAA was evaluated in two medium, homogeneous and heterogeneous. It is important to emphasize that the TPS under TBI conditions was neither fed nor modeled. The objective of this topic is to evaluate its accuracy against experimental data and, as a result, make some decisions that could be clinically advantageous, such as using or not the TPS as a source for monitor unit calculation, for example, and so on. For homogeneous medium, PDDs, TPR (converted, measured and conventional), output factors, dose profiles and monitor unit calculated by TPS were confronted with experimental measurements in differents geometries as described in topic 2.1. The UM calculation involved scanning the phantom shown in figure 1. The reference configuration was used to compare the TPS with experimental data. During this comparison, the SSD was varied from 290 cm to 330 cm. For the heterogeneous medium, a chest simulator from Standard Imaging Middleton, WI USA (IMRT Phantom reference 91230), consisting of acrylic and two slabs with a density similar to a human lung (0.3 g/cm^3) and with different inserts for IC was used (figure 2). Doses were measured for different SSDs in the range of 330 cm to 290 cm and compared with the TPS dose prediction.

Figure 2. *Configurations of the CT imaging and linear accelerator measurements of the acrylic chest phantom.*



3. RESULTS AND DISCUSSION

3.1 Tissue phantom Ratio results

Analyzing the figure 3, from the depth of 5 to 9 cm, for energies of 6 and 10 MV, the converted TPR underestimates the TPR value measured in TBI conditions. And from 10 cm to 29 cm, it overestimates. This overestimation is associated with the lack of correction of the scattering pattern using the Mayneord factor. For 6MV, the deviation between the converted and measured TPR reached a maximum value and an average of (-2.1 ; -1.2) % and (4.0 ; 2.1) %, respectively, in the two regions mentioned above. And for 10 MV, the value are (-1.7 ;-0.9) % and (12 ; 4.3) %, respectively. According to the figure 4, it was clearly observed that the deviation increases with depth. In the case of using converted data, from a clinical perspective, this difference would have more significant implications for the bilateral technique than for the AP/PA technique. On average, if we used the converted data and not those measured under TBI conditions, the statistical uncertainty would represent between (-1.2 to 2.1) % for 6 MV and for 10 MV (-1.7 to 4.3) % over the monitor units. According to Van Dyk et al. [10], for measurements on the central axis such as PDD, TMR or TPR, a deviation between 2 and 6 % is expected. It is essential that converted data be validated in treatment situations.

Now comparing the TPR measured in conventional SSD with that measured in TBI conditions, we observe that for 6 MV, it underestimates between the depths of 5 to 9 cm and between 24 to 29 cm with the maximum and average value of -2.4 % and -1.4 %, respectively. And for other depths it overestimated with a maximum and average of 2.7 % and 1.3%. And for 10 MV, the conventional TPR underestimated from depths of 5 to 9 cm with the maximum and average value of -1.4 and -0.7 % and other points overestimated with maximum and average of 8.0 % and 2.7 %, respectively. It is important to emphasize that points above 20 cm depth were extrapolated from the measured ones, and perhaps this difference of 8 % is associated with backscattering that is not considered to be the result of the extrapolation.

On the other hand, although the TPR is considered independent with distance, there is evidence that for extreme distances relative to the conventional one there is a certain dependence with SSD, according with Van Dyk et al. [10].

If we used the TPR data measured in conventional SSD and not those measured in TBI conditions, the statistical uncertainty about the monitor units would be (±1.4%) for 6 MV and (-1.4 to 2.7) % for 10 MV. The authors of this study recommend using the experimental data under TBI conditions.

Figure 3 *Comparison between measured, converted and conventional TPR for 6 MV*

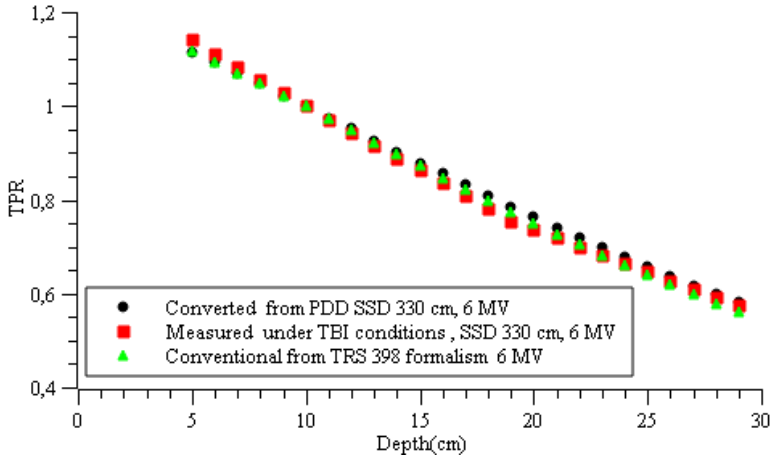
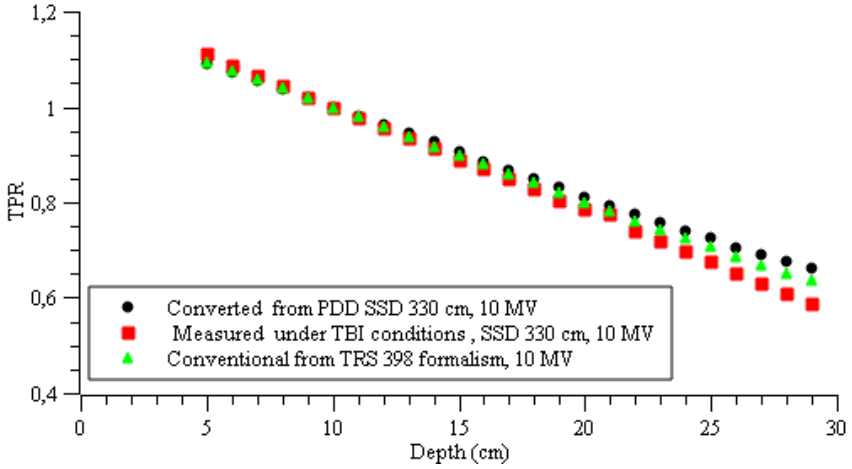


Figure 4 *Comparison between measured, converted and conventional TPR for 10 MV*



3.2 Evaluating the dose variation with distance

The ISL measurements are usually made in air and are, however, affected by the scattering from the collimators, floor and walls of the room. The deviation should be ≤ 2 % according to Van Dyk et al. [10]. A maximum deviation of -1.8 % was found in the present study. Table 2 and 3 shows the results obtained for the 6 and 10 MV beams.

3.3 Effect of irradiated cable

The presence of 2 m of irradiated cable did not produce a variation in the reading of more than 1 % for 6 MV and 10 MV energy. The results are within the expected according to Van Dyk et al [10] which points to an increase between 1 and 2 % of the ionization current when 2 m of cable is irradiated.

3.4 Dose rate constant test

Results of this investigation found a linearity of the dose with the dose rate of ± 0.5 %. These results agree with that reported by Klein et al. [11], which suggests that the variation of the rate with dose should be ± 2 %.

Table 1 Test of the inverse square Law, 6 MV

SSD (cm)	Fdist_calculated	Fdist_Measured	Deviation (%)
336	0.966	0.966	0.00
341	0.938	0.944	0.63
327	1.018	1.010	-0.79
316	1.088	1.073	-1.37
310	1.129	1.109	-1.77

Table 2 Test of the inverse square Law, 10 MV

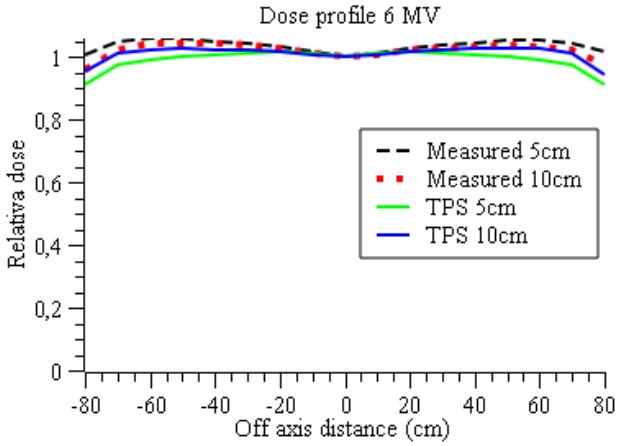
SSD (cm)	Fdist_calculated	Fdist_Measured	Deviation (%)
336	0.966	0.966	0.00
341	0.938	0.943	0.53
327	1.018	1.010	-0.78
316	1.088	1.074	-1.28
310	1.129	1.111	-1.59

3.5 TPS Perfomance Check

3.5.1 Dose Profile

The figure 5 shows the comparison between the measured and the calculated dose profiles. For 6 MV beam measured, 0.2 % symmetry and 4.3 % flatness was found and for TPS profile 0.3 % and 2.9 % of symmetry and flatness, respectively was calculated at 10 cm depth. Cananoglu et al. [14], for SSD = 350 cm, measured a flatness of 4.79 % and observed that beam flatness decreased with increasing SSD. The tolerance presented by Van Dyk et al. [10] is 3 % for a 30 cm x 30 cm x 30 cm phantom.

Figure 5 : *Dose profile comparison between measured and TPS calculated data.*



3.5.2 Percentage depth dose results

Figure 6: *Comparison of PDD with and without spoiler for 6 MV*

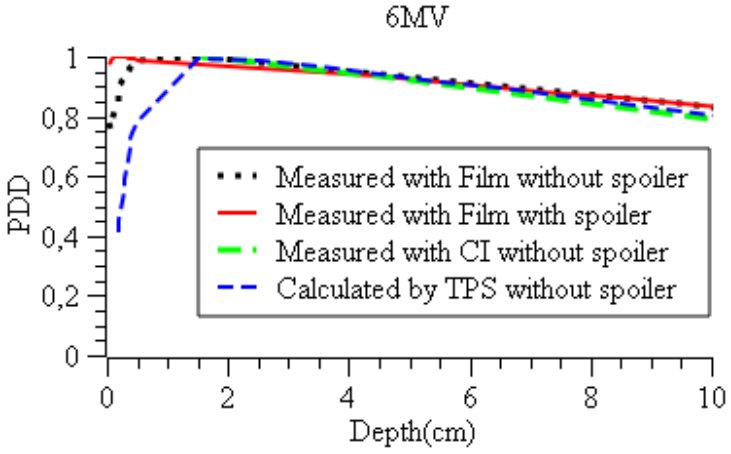
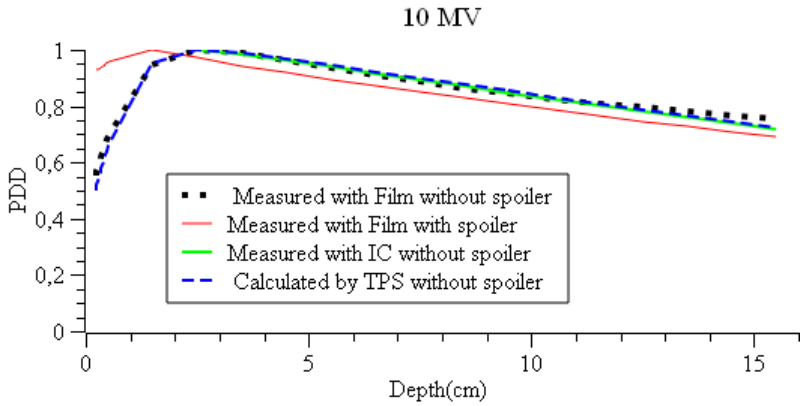


Figure 7: *Comparison of PDD with and without spoiler for 10 MV*



For the 6 MV beam, the entrance dose measured with film was 75 % of the maximum dose without a spoiler and 99.5 % with a spoiler (figure 6). For the 10 MV beam without a spoiler, the entrance dose was 55 %, and 93 % with the spoiler (figure 6). These findings demonstrate the usefulness of the spoiler for clinical purposes, which is consistent with the results reported by Zeng Q. [15] and Khan et al. [5].

Figure 6 shows that there is a tendency to underestimate the measured dose using IC as compared to film, as the depth increases. On average, this deviation reaches -4.5 % after 1.5 cm. It is crucial to emphasize that due to restrictions in the configuration, the PDD measured with IC was obtained starting from a depth of 1.5 cm and with a minimum interval of 1 cm. The TPS in general shows a tendency to overestimate relatively to the IC with mean deviation of 1.5 % and an underestimation to film with mean deviation of -3.5 %. At depths below 1.5 cm the underestimation reached up to -20 %.The dose in the Build Up region, depends on many factors, as field size, energy, etc. In TBI scenario, longer SSD and an intervening air could be associated to this difference. Probably, the calculation algorithm, did not correct the scattering pattern and other products resulting from the extended SSD and the an intervening air.

For the 10 MV beam, the figure 7 shows that the TPS exhibits a strong correlation with both IC and film measurements, with only a small average deviation of around ± 1 %. Between depths of 5 cm and 9.5 cm, the IC overestimates by an average of 0.5%, while at other depths, it underestimates by an average less than -2 % when compared to the film measurement. Lamichhane et al. [12] also reported a tendency for underestimation more than 5 % for PDD measurements under TBI conditions.

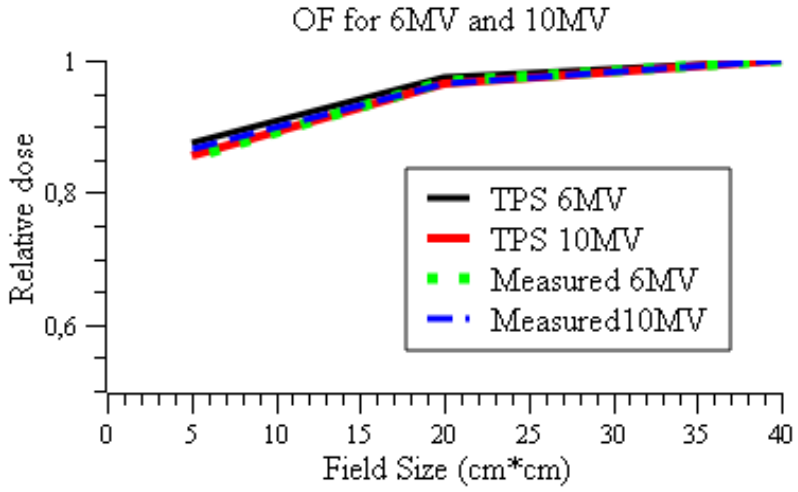
3.5.3 Monitor Unit calculation

When determining the MU for a SSD ranging from 290 cm to 330 cm, it was noted that the TPS tended to overestimate the MU values by 2.7 % for a 6 MV and underestimate them by -2.6 % for a 10 MV. Although deviations of this order (± 2.7 %) can be considered reasonable for the TBI scenario, the authors of this study chose not to use the MU values provided by the TPS. This precaution was adopted because it was understood that for the clinical implementation of TPS for TBI, a more comprehensive evaluation should be conducted, covering different scenarios. Furthermore, the planning system was not calibrated for TBI conditions. As a result, scattering, backscattering, and other phenomena arising from beam interactions were not taken into account. This introduces additional uncertainties in the MU calculation process. Other researchers [12,13] also recommend that the use of TPS should be done with caveats. TPS can be used as a secondary check of the manual calculation made from the experimental data or to evaluate the relative dose distribution in a three-dimensional planning.

3.5.4 Output Factor Measurements

Regarding the output factors, the largest discrepancy observed between TPS and experimental measurement was for 5 x 5 cm^2 field, with deviations of was 2.4 % and -0.9 % for 6 and 10 MV, respectively as shown in figure 7. According Klein et al. is expected ± 2 %. Probably the correction of scattering pattern could be associated with this small difference.

Figure 8 *Output factor comparison between Measured and calculated data for 6 and 10MV*



According to the results, the authors of the present article suggest using the dose rate treatment between 5 to 10 cGy/min, as recommended by Van Dyk et al. And for calculating the monitor unit, the following equation is suggested:

$$MU = \frac{Dose\ prescrita}{F_{C\ Linac} \times F_{C\ TBI} \times TPR_{LM} \times \left(\frac{SSD_{cal}+d_{ref}}{SSD+L_M}\right)^2} \quad (3)$$

Where $F_{C\ Linac}$ and $F_{C\ TBI}$ the calibration factors measured at standard and extended SSD respectively and TPR_{LM} is the TPR value at midline (L_M).

$$L_M = 0.5 \times DAP_{média} \quad (4)$$

3.5.5 Heterogeneity

The TPS underestimated dose values by up to - 3.5 % with a mean deviation of – 2.9 %, for 6 MV. For 10 MV, the TPS overestimated the dose values by up to 1.09 %, with an average deviation of 1.01 % using the acrylic thorax phantom. Tables 3 and 4 summarize the results obtained.

Table 3 Comparison between TPS and measured dose for 6 MV in the acrylic phantom.

SSD (cm)	TPS Dose(cGy)	Measured Dose (cGy)	Diff. (%)
(330±0.05)	(11.9±0.1)	(12.24±0.02)	-2.8
(320±0.05)	(12.6±0.1)	(13.06±0.01)	-3.5
(310±0.05)	(13.4±0.1)	(13.83±0.01)	-3.1
(300±0.05)	(14.3±0.1)	(14.70±0.01)	-2.7
(290±0.05)	(15.3±0.1)	(15.69±0.01)	-2.5

Table 4 Comparison between TPS and measured dose for 10 MV in the acrylic phantom

SSD (cm)	TPS Dose(cGy)	Measured Dose (cGy)	Diff. (%)
(330±0.05)	(12.5±0.1)	(12.36±0.01)	1.09
(320±0.05)	(13.3±0.1)	(13.17±0.01)	0.98
(310±0.05)	(14.1±0.1)	(13.97±0.03)	0.95
(300±0.05)	(15.0±0.2)	(14.84±0.03)	1.07
(290±0.05)	(16.0±0.2)	(15.84±0.01)	0.99

4. CONCLUSIONS

In this study, the 6 and 10 MV photon beams of the Varian accelerator CX model were commissioned under TBI conditions for treatment. A set of data were acquired, highlighting the TPR table and methodology for manually calculating the monitor unit. The experimental data also, were confronted with those calculated by TPS in different clinically relevant configurations either in homogeneous or heterogeneous medium and additionally with data converted from measurements in conventional SSD. It was observed an increase in deviation with increasing depth and with heterogeneity between the experimental data and the TPS. Regarding the dose, the TPS showed a tendency of underestimation by approximately -3.5 % or less and overestimation by approximately 1.09 % or less for 6 and 10 MV, respectively compared to the experimental values.

The use of the spoiler was shown to be clinically advantageous where, for a 6 MV beam, the entrance Percentage Depth Dose was 75 % without a spoiler and 99.5 % with a spoiler. For a 10 MV beam, it was verified that without a spoiler, the entrance PDD was about 55 %, but with a spoiler, it was about 93 % . Although the TPS shows a deviation of ± 2.7 % from the experimental data in terms of MU calculation, it will not be used as a source for MU calculation, but rather as a secondary check and as a tool to evaluate the dose distribution. Results of this study are based on a specific set up.

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