RELATIVE OUTPUT FACTORS OF DIFFERENT COLLIMATION SYSTEMS IN TRUEBEAM STX MEDICAL LINEAR ACCELERATOR

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Purpose: Determine relative output factors of differrent collimation systems of a medical linear accelerator to ensure better modelling of the photon beam, especially small fields, as used in radiotherapy treatment planning systems; **Method:** The IAEA TRS483 and TRS398 Code of Practices (CoP) were used to calculate relative output factors for photon beams of 6X, 6XFFF energies with High Definition Multileaf Collimator (HDMLC), jaws and cones mounted on TrueBeam STx medical linear accelerator (Varian Medical Systems). A comparison between these results were made to show physicical characteristics of different collimation systems; **Result**: There is a large discrepancy in relative output factor curves found among different collimation systems of the same equivalent field sizes and between the CoPs. **Conclusion**: Specific beam modelling for each type of collimation system maybe required in TPS for better computation accuracy.

Keywords: TRS483 code of practice, small field dosimetry, relative output factors.

1. INTRODUCTION

Modern radiotherapy techniques such as Intensity-Modulated Radiation Therapy (IMRT), Volumetric Modulated Arc Therapy (VMAT), Stereotactic Radiosurgery (SRS) and Stereotactic Radiation Therapy (SRT) make use of small photon beams in order to deliver complex radiation treatments. However, there are still many physical and technical aspects which need to be considered in order to commission small photon beams safely and efficiently into clinical practice such as: changing in photon fluence spectrum making beam quality changing by field size, lateral disequilibrium of charged particles may leading to wrong estimation of absorbed dose as well as detector size compared to field size [1]–[4].

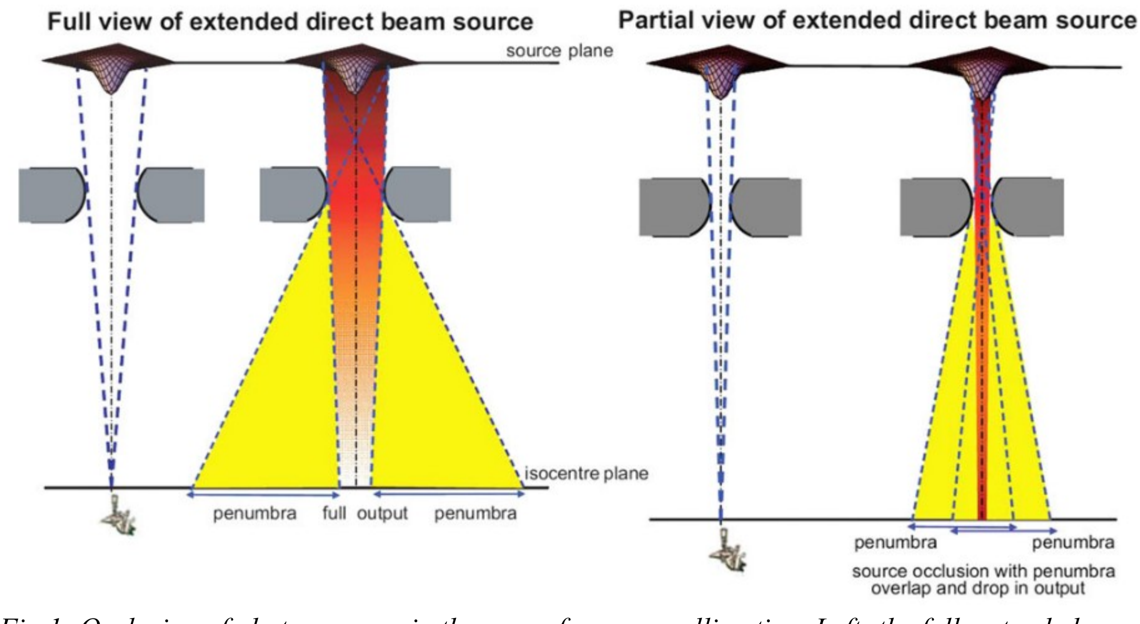


Fig.1: Occlusion of photon source in the case of narrow collimation. Left: the full, extended source can be "viewed" by an observer on the central axis. Right: only partial view of the source is possible by an observer on the central axis. (Adapted from IPEM Report 103, Aspradakis et al.).

Practical issues encountered are: photon beam data for treatment planning system (TPS) are usually collected for jaw-shaped beams while we use these data for computation of MLC-shaped beams. Furthermore, HDMLC-shaped beams are constituted from very tiny beamlets, much smaller than smalest collected beam data of field size 3 × 3 cm^2 (at isocenter), which may affect the computation accuracy of TPS, especially for small tumors. In Eclipse v.13.6 (Varian Medical Systems), warning message “inaccuracy” was often seen when making treatment plans for tumors less than 3cm diameter. Radiation oncologists tend to use HDMLC for small tumor radiosurgery because of its small thickness (2.5 mm at isocenter) and convenience.

Cone collimators are dedicated for radiosurgery of small tumors. With cone-shaped beams, field size diameters are of 17.5 mm down to 4 mm cone but they are previously measured using TRS398 CoP (IAEA). It has been shown that the beam quality of photon beam changes significantly due to these very small field collimations [2], [5]–[9]. In this study, we made a comparison of relative output factors of different collimation systems (jaws, HDMLC and cones) for further estimation of computation accuracy of TrueBeamSTx TPS using newly published TRS483 CoP (IAEA).

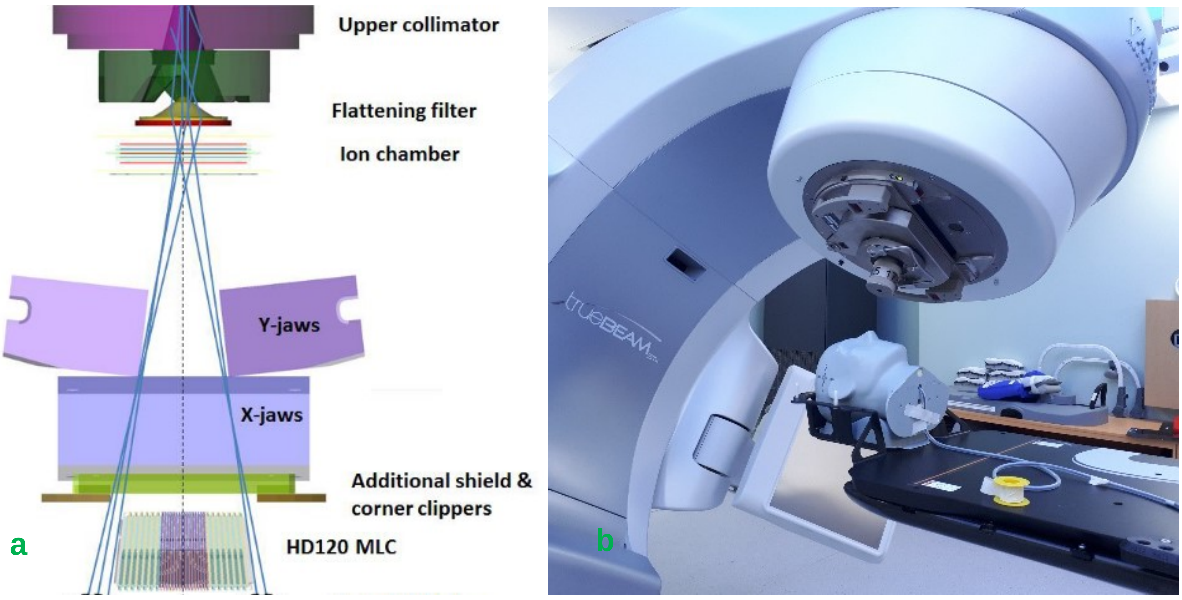


Fig.2: TrueBeam STx treatment head diagram with collimation systems: a) Jaws (highest), HDMLC (midlle) [10] and b) Cone (lowest)

2. MATERIAL AND METHOD

TrueBeamSTx (Varrian Medical Systems) medical linear accelerator with integrated HDMLC (20 central leaf pairs of 2.5mm thickness and 40 peripheral leaf pairs of 5.0mm thickness at isocenter). Beam shaping using High Definition MLC (and also jaws) were of field sizes 0.5 × 0.5, 1 × 1, 2 × 2, 3 × 3, 4 × 4, 5 × 5, 7 × 7, 10 × 10 cm^2. MLC-shaped field were created when jaws were “optimized” and at “recommended positions” by software. Inversely, jaw-shaped field were created when MLC are fully retracted. Beam shaping using the cones are with diameter of 4.0, 5.0, 7.5, 10.0, 12.5, 15.0 and 17.5 mm. Photon energies of 6X (with Flatterning Filter), 6XFFF (Flatterning Filter-Free), 10X, 10XFFF were used for measurements. The linac was calibrated for all photon energies at 10 × 10 cm^2 jaw-shaped field to be used for all other collimation systems.

The dose measurements were performed in Blue Phantom 2 (IBA) using a Razor chamber (IBA) and Razor diode (IBA) under Source-to-Axis Distance setup (100 cm SAD, 5 cm depth). The TRS398 and TRS483 CoP are both applied to determine relative output factors. Relative output factor curves were compared for 3 different collimation systems and in both CoPs. All data were normalized to 10× 10 cm^2 field size. The equivalent square of the cone defined fields were calculated using formular [2] :

$$S_{clinical} = \sqrt{A \times B} = r\sqrt{\pi} \quad (1)$$

Razor diode (unshielded, p-type silicon diode chip, active detector diameter of 0.6 mm) with high spatial resolution and high sensitivity is superior to Razor chamber (total active length of 3.6mm) in relative dosimetry of small photon beams. However, Razor diode has an over-response in large fields because of the significant amount of phantom scatter component of low energy photons. The consequence is an underestimation of field output factors when they are normalized to a large field size (e.g. the conventional 10 cm × 10 cm^2 reference field) [2].

According to TRS398 CoP, the output factor may be determined as the ratio of corrected dosimeter readings measured under a given set of non-reference conditions to that measured under reference conditions. However, in TRS483 CoP, the field output factor, $\Omega_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$, relative to f_{msr} is defined by:

$$\Omega_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}} = \frac{M_{Q_{clin}}^{f_{clin}}}{M_{Q_{msr}}^{f_{msr}}} k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}} \quad (2)$$

where $M_{Q_{clin}}^{f_{clin}}$ and $M_{Q_{msr}}^{f_{msr}}$ are the readings of the detector (corrected for influence quantities) in the clinical field (f_{clin}) and the machine specific reference field (f_{msr}), respectively. $k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ is beam quality correction factor which changed by field size.

The intermediate field (f_{int}) method were used with two detectors the Razor ionization chamber and the Razor unshielded diode.

$$\Omega_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}} = \left[\frac{M_{Q_{clin}}^{f_{clin}}}{M_{Q_{int}}^{f_{int}}} k_{Q_{clin},Q_{int}}^{f_{clin},f_{int}}\right]_{det} \left[\frac{M_{Q_{int}}^{f_{int}}}{M_{Q_{msr}}^{f_{msr}}} k_{Q_{int},Q_{msr}}^{f_{int},f_{msr}}\right]_{IC} \quad (3)$$

where “det” refers to the small field detector (Razor diode) and “IC” to the ionization chamber (Razor chamber). The output correction factor $\left[k_{Q_{clin},Q_{int}}^{f_{clin},f_{int}}\right]_{det}$ is obtained from the tabulated output correction factors with respect to the machine specific reference field as below:

$$\left[k_{Q_{clin},Q_{int}}^{f_{clin},f_{int}}\right]_{det} = \frac{\left[k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}\right]_{det}}{\left[k_{Q_{int},Q_{msr}}^{f_{int},f_{msr}}\right]_{det}} \quad (4)$$

3. RESULTS

3.1. Output factors of collimation systems using TRS398 CoP:

Using conventional formular for output factor analysis (TRS398 CoP), we got the result as Table 1.

Table 1. Output factors of collimation systems using TRS398 CoP and Razor chamber.

Cone (mm)	**4**	**5**	**7.5**	**10**	**12.5**	**15**	**17.5**	**100 (square)**
6X	0.420	0.523	0.662	0.746	0.797	0.834	0.859	1
6XFFF	0.473	0.574	0.702	0.772	0.816	0.846	0.866	1
MLC Field size	**0.5**	**1**	**2**	**3**	**4**	**5**	**7**	**10**
6X	0.529	0.754	0.861	0.895	0.921	0.940	0.968	1
6XFFF	0.560	0.782	0.876	0.909	0.932	0.948	0.975	1
Jaw Field size	**0.5**	**1**	**2**	**3**	**4**	**5**	**7**	**10**

6X	0.345	0.704	0.850	0.888	0.914	0.936	0.967	1
6XFFF	0.377	0.736	0.867	0.906	0.929	0.947	0.974	1

3.2. Output factor of collimation systems using TRS483 CoP:

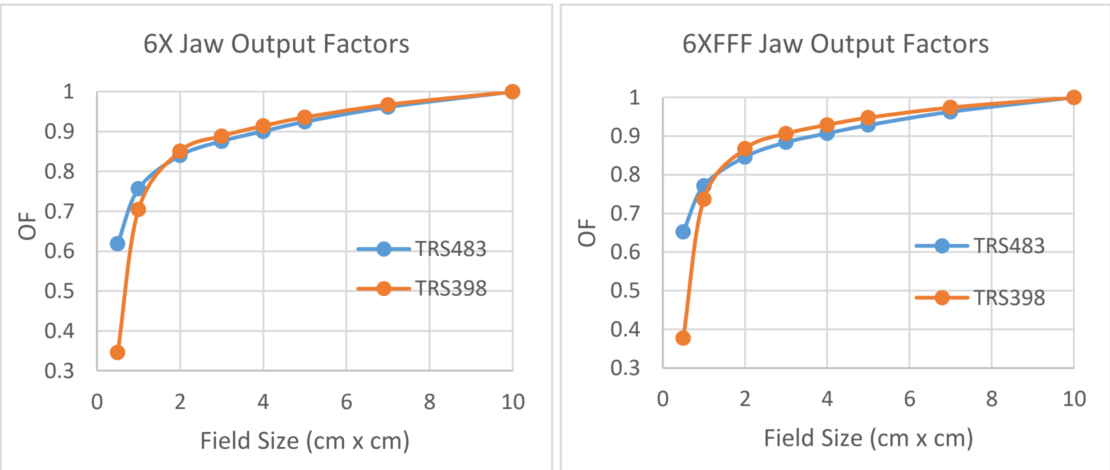
Intermediate field (f_{int}) of 4 × 4 cm^2 was selected for calculation of jaw-shaped fields and MLC-shaped fields. For cone, intermediate field was 17.5mm conical field because we need to normalize these data to that of 10 × 10 cm^2 field size. The results were obtained as Table 2.

Table 2: Output factor of collimation systems using TRS483 CoP (Razor chamber and Razor diode)

Cone (mm)	**4**	**5**	**7.5**	**10**	**12.5**	**15**	**17.5**	**100 (square)**
Square Equi. Field size (cm)	**0.708**	**0.885**	**1.327**	**1.77**	**2.212**	**2.655**	**3.097**	**10**
6X	0.522	0.599	0.713	0.773	0.814	0.841	0.864	1
6XFFF	0.578	0.648	0.745	0.797	0.830	0.856	0.871	1
MLC Field size	**0.5**	**1**	**2**	**3**	**4**	**5**	**7**	**10**
6X	0.608	0.774	0.871	0.905	0.927	0.945	0.971	1
6XFFF	0.643	0.792	0.881	0.918	0.938	0.954	0.978	1
Jaw Field size	**0.5**	**1**	**2**	**3**	**4**	**5**	**7**	**10**
6X	0.619	0.756	0.84	0.876	0.901	0.924	0.961	1
6XFFF	0.652	0.771	0.846	0.884	0.907	0.928	0.963	1

4. DISCUSSION

4.1. Comparison of results between TRS483 and TRS398 CoP:



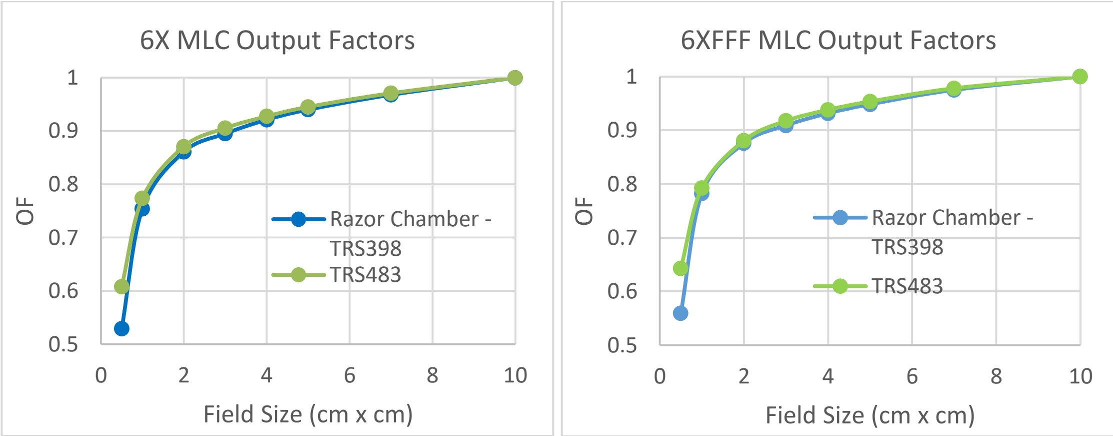


Fig.3: Difference of TRS398 and TRS483 CoP in relative output factor of MLC and Jaws collimations.

Based on these results, the difference between ROF curves is significant between the two different methods (CoPs) for MLC-shaped field size and for jaw-shaped field size less than 3 × 3 cm for both the 6X and 6XFFF beams.

The smallest difference was observed with MLC-shaped fields while the biggest difference was observed with cone-shaped fields as seen in Fig.3 and Fig.4. At 0.5 × 0.5 cm^2 squared field and 4 mm conical field, the output factor difference of 6X and 6XFFF beams were -44.2%/-42.2%, -13.0%/-13.0%, -19.6%/-18.2% for jaw-shaped, MLC-shaped and cone-shaped fields, respectively. TRS398 CoP gave underestimation of relative output factor in comparison with TRS483 CoP. Large difference were always seen at field sizes smaller than 4 × 4 cm^2.

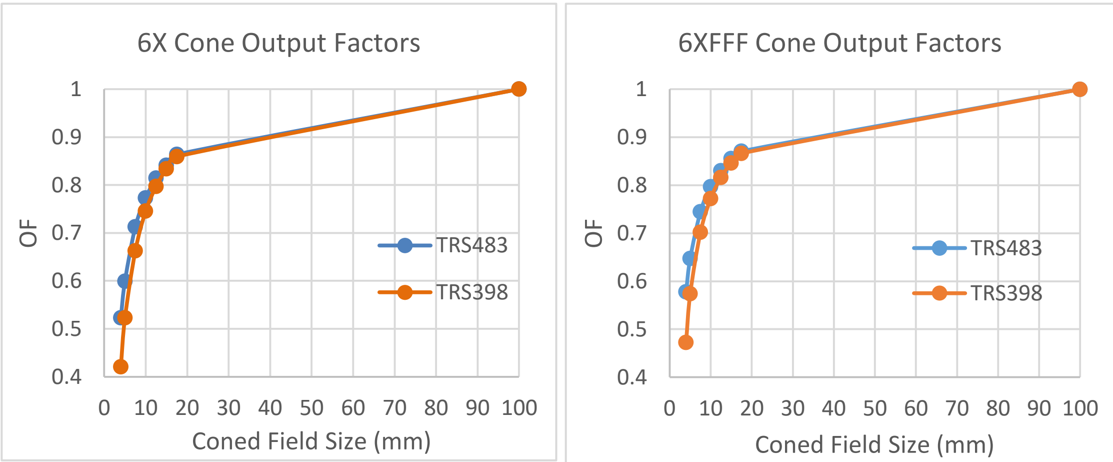


Fig.4: Difference of TRS398 and TRS483 CoP in relative output factor of cone collimations.

Noticingly, the Razor chamber's reading differences between 10 × 10 cm^2 MLC-shaped field and 10 × 10 cm^2 jaw-shaped field were just 0.61% and 0.25% for 6X and 6XFFF, respectively. Therefore, these relative output factor could be used for direct comparison between jaw-shaped field and MLC-shaped field of "the same" nominal field size.

4.2. Comparison of results between 6X, 6XFFF (TRS483 CoP):

For the same collimation system, output factor comparisons were also made for 6X and 6XFFF beams after applying TRS483 CoP. The biggest differences in output factor were seen at 0.5 × 0.5 cm^2 jaw-shaped field, 0.5 × 0.5 cm^2 MLC-shaped field and 4mm cone-shaped field with values of 5.3%, 5.8% and 10.5%, respectively.

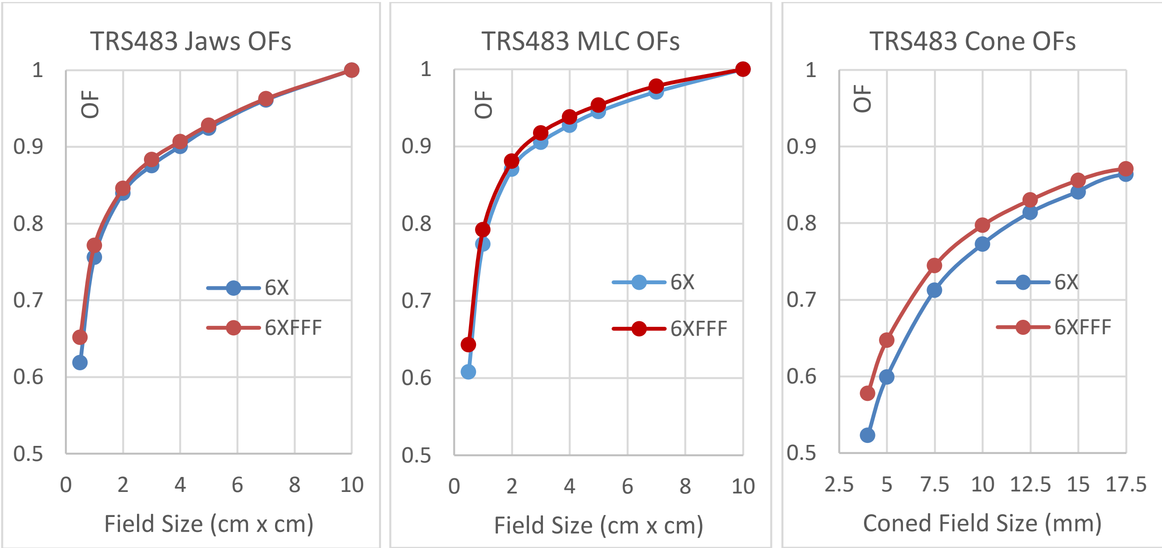


Fig.5: Difference in output factor of 6X and 6XFFF beams in each collimation system.

4.3. Comparison of Output Factor curves between different collimation systems (TRS483 CoP):

The relative output factor comparisons were made between MLC, jaws and Cone systems for both 6X and 6XFFF beams.

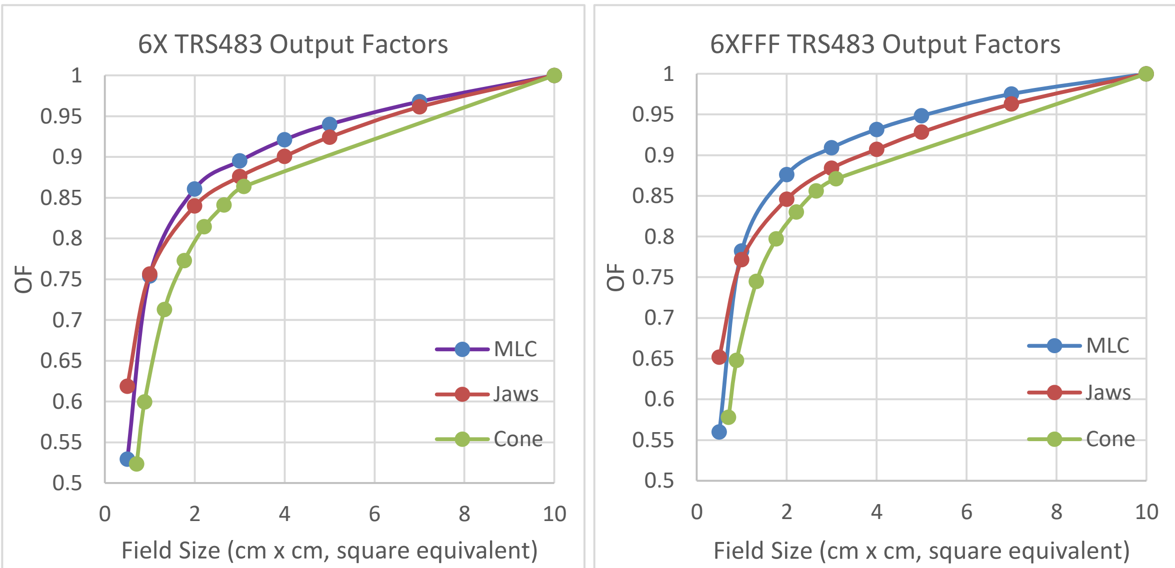


Fig.6: Difference in output factor of difference collimation system for 6X and 6XFFF beams.

Conical collimators are independent from MLC and Jaws systems. Conical collimation system has smallest relative output factor in comparison with that of MLC and Jaws systems for both 6X and 6XFFF beams as Fig.6.

For field sizes bigger than 1×1 cm^2, jaw system has lower relative output factor than MLC's but it is inverse for field size less than 1×1 cm^2.

In a multi-centre analytical study of small field output factor calculations in radiotherapy reported by Krzysztof Chełmiński and Wojciech Bulski, for 2×2 cm^2 MLC-shaped fields of Varian linacs, the differences between the treatment planning system output factors (based on collected beam data) often exceeded 5% and were below 10% [11]. In our study, these differences were -1.1% (6X) and -0.5% (6XFFF) for MLC-shaped fields, 1.3% (6X) and 2.6% (6XFFF) for jaw-shaped fields, -7.0% (6X) and -5.7% (6XFFF) for cone-shaped fields. The smaller differences observed in our study for MLC-shaped field may came from our small field detector, the Razor chamber.

A multinational audit of small field output factors calculated by treatment planning systems used in radiotherapy, the OFs for small fields calculated by TPSs were generally larger than measured reference data. On a national level, 30% and 31% of the calculated OFs of the $2 \times 2\ cm^2$ field exceeded the action limit of 3% for nominal beam energies of 6 MV and for nominal beam energies higher than 6 MV, respectively [12].

The discrepancy above may come from accuracy of treatment planning algorithms on measured output factors, especially for small fields.

CONCLUSION

TRS483 CoP was successfully applied to recalculate relative output factors for cone system with correction. Relative output factors for jaw collimation system were extensively obtained for field size less than $3 \times 3\ cm^2$ for Eclipse v.13.6 for 6X and 6XFFF beams using TRS483 CoP. Relative output factors were also measured for MLC collimation system to be compared with that of jaw collimation system. The discrepancy of output factor between jaw-shaped fields and MLC-shaped fields suggests that jaw-based beam data itself may not suitable for MLC-based treatment planning. Additional measurement of small beam percentage depth dose and profiles as well as specific modelling of photon beam for MLC system may be required.

REFERENCES

[1] P. Andreo, "The physics of small megavoltage photon beam dosimetry," *Radiother. Oncol.*, vol. 126, no. 2, pp. 205–213, 2018.

[2] International Atomic Energy Agency, TRS483, "An International Code of Practice for Reference and Relative Dose Determination", November, 2017.

[3] I. J. Das, G. X. Ding, and A. Ahnesjö, "Small fields: Nonequilibrium radiation dosimetry," *Med. Phys.*, vol. 35, no. 1, pp. 206–215, 2008.

[4] I. J. Das, J. Morales, and P. Francescon, "Small field dosimetry: What have we learnt?," in *AIP Conference Proceedings*, 2016.

[5] K. Park, J. Bak, S. Park, W. Choi, and S. W. Park, "Determination of small-field correction factors for cylindrical ionization chambers using a semiempirical method," *Phys. Med. Biol.*, vol. 61, no. 3, 2016.

[6] H. Bouchard, J. Seuntjens, S. Duane, Y. Kamio, and H. Palmans, "Detector dose response in megavoltage small photon beams. I. Theoretical concepts," *Med. Phys.*, vol. 42, no. 10, 2015.

[7] E. Mainegra-Hing and B. R. Muir, "On the impact of ICRU report 90 recommendations on kQ factors for high-energy photon beams," *Med. Phys.*, vol. 45, no. 8, pp. 3904–3908, 2018.

[8] H. Benmakhlouf, J. Sempau, and P. Andreo, "Output correction factors for nine small field detectors in 6 MV radiation therapy photon beams: A PENELOPE monte carlo study," *Med. Phys.*, vol. 41, no. 4, 2014.

[9] I. J. Das, M. B. Downes, A. Kassaee, and Z. Tochner, "Choice of Radiation Detector in

Dosimetry of Stereotactic Radiosurgery-Radiotherapy," *J. Radiosurgery*, vol. 3, no. 4, pp. 177–186, 2000.

[10] M. Constantin, H. Hsu, S. Mansfield, and M. Svatos, "Mitigation Strategies to Reduce the Relative Extrafocal Dose (RED)," vol. 33, p. 2525, 2006.

[11] K. Chełmiński and W. Bulski, "A multi-centre analytical study of small field output factor calculations in radiotherapy," *Phys. Imaging Radiat. Oncol.*, vol. 6, no. September 2017, pp. 1–4, 2018.

[12] W. Lechner *et al.*, "A multinational audit of small field output factors calculated by treatment planning systems used in radiotherapy," *Phys. Imaging Radiat. Oncol.*, vol. 5, no. September 2017, pp. 58–63, 2018.