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Comparision of beam data requirements for MLC commissioning on a TPS

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The treatment planning system (TPS) has become a key element in the radiotherapy process with the introduction of computer tomography (CT) based 3D conformal treatment planning. Commissioning of a MLC on a TPS either for conformal radiotherapy or intensity modulated radiation therapy (IMRT) requires beam data to be generated on a linear accelerator. Most of the TPS require these beam data to be generated with routine collimator jaws. However some TPS demand the data to be provided for MLC shaped fields. This prompted us to investigate whether beam data with jaws differ than that with MLC and whether the jaw based beam data would suffice for the commissioning of a MLC on a TPS.

Beam data like percentage depth dose (PDD), cross beam profiles and output factors was acquired for jaws and MLC defined square fields for 6, 10 and 23 MV photon beams. Percentage depth dose and cross beam profiles were acquired with a radiation field analyzer RFA-200, CC13-S ion chambers with active volume of 0.13 cm³ and OmniPro-Accept software from Scanditronix-Wellhofer. A Medtec-TG51 water tank with Max-4000 electrometer and 0.6 cc PTW ionization chamber and a mini phantom from Standard Imaging was utilized for output measurements for millennium-120 MLC (Varian Medical Systems) and SRS diode detector (Scanditronix-Wellhofer) of 0.6 mm diameter of active area and 0.3 mm of active volume thickness for micro-MLC (BrainLab).

The difference in PDD in the build-up region for millennium MLC was $\pm 1.0\%$ for 6 MV photons. For 10 MV photons the PDD difference was within $\pm 4.0\%$. The difference in PDD for 23 MV photons ranged from 0% to 40.0%. PDD difference from build-up depth to about 28 cm was within $\pm 1.0\%$. Difference in PDD crossed $\pm 1.0\%$ at 30 cm depth for 6 MV photons. The difference in PDD in the

build-up region for mMLC was $\pm 8.0\%$ for 6 MV photons. For the smallest field size studied with micro-MLC i.e. $0.6 \times 0.6 \text{ cm}^2$ difference in PDD was more than $\pm 1.0\%$ in the build-up region and beyond a depth of 8.0 cm. The profiles for jaws and MLC agreed within the umbra region. However in the penumbra region small differences in doses were observed. The collimator scatter factor (S_c), phantom scatter factor (S_p) and output factor values for MLC were different than those for jaws.

The differences in beam characteristics could have implication for intensity modulated radiation therapy and stereotactic radiosurgery in terms of dose in the build up region, exit dose, dose to the planning target volume (PTV) and organ at risk (OAR). Impact of these dosimetric differences between jaw and MLC needs to be further studied in terms of dose volume histograms for PTV and OAR and its further impact on tumor control probability (TCP) and normal tissue complication probability (NTCP).

Key words: beam data, MLC commissioning, TPS.

Introduction

A computerised treatment planning system (TPS) is an essential tool in the design of a radiotherapeutic treatment of cancer patients and some groups of patients suffering from non-malignant diseases [3, 5, 7, 8]. Treatment planning systems have provided the radiation therapy community with an important tool for more than four decades and must receive some of the credit for the advances made in treatment outcome during this period. This task that TPS were applied to was the calculation of a dose distribution, which albeit important is only one component of the overall task of planning the therapy of cancer patients. In recent years advances in medical imaging have made available the data necessary to greatly improve the patient model used for treatment planning. This development in combination with graphic displays and the tremendous decrease in cost of computer capability spurred the development of TPS that addressed a much broader range of problems than just dose calculations, such systems are called 3D or multidimensional, which provide the capability namely synthesis of all relevant diagnostic information in evaluating the patient's disease, appreciation and delineation of anatomy, simulation of therapy, design of treatment aids, calculation of 3D dose distributions, evaluation of treatment plan, verification of treatment and follow-up studies. The treatment planning system (TPS) has become a key element in the radiotherapy process with the introduction of computer tomography (CT) based 3D conformal treatment planning.

Multileaf Collimators (MLC) are computer-controlled devices that are capable of providing photon beam shielding for linear accelerators using high density leaves

(typically tungsten alloy) which are projected into the radiation field. Multileaf Collimators have a number of uses in radiation therapy, the primary being to replace conventional mounted blocks for critical organ shielding. When used in this manner MLC's eliminate the need for mounted block manipulation and storage, and the computer controlled nature of the device means that changes in the blocking pattern can be easily produced. A secondary use of MLC's is in the production of intensity modulated beams. By adjusting the position of the MLC in the radiation field during the beam-on time, an arbitrary dose distribution may be achieved. Current MLC systems vary with respect to design, location and use. They may be installed as a tertiary device below the secondary collimators, or they may comprise a total or partial replacement of the secondary collimators. The leaves must provide an acceptable degree of beam attenuation, provide a large enough field coverage, and must be well integrated with the rest of the collimator shaping system. In order to minimize penumbra various design considerations have been devised by manufacturers to provide focused field shaping.

Commissioning of a MLC on a TPS either for conformal radiotherapy or intensity modulated radiation therapy (IMRT) requires beam data to be generated on a linear accelerator namely percentage depth dose (PDD), beam profiles, output factors and MLC transmission. Most of the TPS require these beam data to be generated with routine collimator jaws. However some TPS like Corvus inverse planning system from Best NOMOS and Brainscan from BrianLab demand the data to be provided for MLC shaped fields. This prompted us to investigate whether beam data with jaws differ than that with MLC and whether the jaw based beam data would suffice for the commissioning of a MLC on a TPS.

Materials and methods

Beam data like percentage depth dose (PDD), cross beam profiles and output factors was acquired for jaws and MLC (millennium 120 MLC) defined square fields on Clinac-iX linear accelerator (Varian Medical Systems) for 6 and 10 MV photon beams, on Clinac-Trilogy linear accelerator (Varian Medical Systems) for 23 MV photon beam and for jaws and MLC (52 leaf microMLC-Brain Lab) defined square fields on Clinac-600C linear accelerator (Varian Medical Systems) for 6 MV photon beam.

Percentage depth dose and cross beam profiles were acquired with a radiation field analyzer RFA-200 with CC13-S ion chambers and OmniPro-Accept software

(Scanditronix Wellhofer). A Medtec-TG51 water tank with Max-4000 electrometer (Standard Imaging) and 0.6 cc PTW ionization chamber and a mini phantom (Standard Imaging) was utilized for output measurements for millennium-120 MLC (Varian Medical Systems) and SRS diode detectors (Scanditronix Wellhofer) for micro-MLC.

Results

The PDD difference between jaw and MLC is shown in Figures 1, 2, and 3 for 6, 10 and 23 MV photon beams for millennium 120 leaf MLC respectively. The difference in PDD in the build-up region was $\pm 1.0\%$ for 6 MV photons. For 10 MV photons the PDD difference was within $\pm 4.0\%$. The difference in PDD for 23 MV photons ranged from 0% to 40.0%. PDD difference from build-up depth to about 28 cm was within $\pm 1.0\%$. Difference in PDD crossed $\pm 1.0\%$ at 30 cm depth for 6 MV photons. The PDD difference between jaw

Figure 1. Difference in PDD between jaw and MLC for 6 MV photons

Figure 2. Difference in PDD between jaw and MLC for 10 MV photons

Figure 3. Difference in PDD between jaw and MLC for 23 MV photons

Figure 4. Difference in PDD between jaw and micro-MLC for 6 MV photons

Figure 5. Profiles for jaw and MLC for 6 MV photons

Figure 6. Profiles for jaw and MLC for 10 MV photons

Figure 7. Profiles for jaw and MLC for 23 MV photons

Figure 8. Profiles for jaw and micro-MLC for 6 MV photons

and MLC is shown in Figure 4 for 6 MV photon beam for micro-MLC. The difference in PDD in the build-up region was $\pm 8.0\%$ for 6 MV photons. For the smallest field size studied with micro-MLC i.e. $0.6 \times 0.6 \text{ cm}^2$ difference in PDD was more than $\pm 1.0\%$ in the build-up region and beyond a depth of 8.0 cm.

The cross beam profiles are shown in Figures 5, 6 and 7 for jaw-millennium MLC combination for 6, 10 and 23 MV photons and in Figure 8 for jaw-mMLC combination for 6 MV photons. The profiles for jaws and MLC agreed within the umbra region. However in the penumbra region small differences in doses were observed. The data regarding differences in output factors is given in Tables 1, 2 and 3 for jaw-millennium MLC combination for 6, 10 and 23 MV photons. The collimator scatter factor, phantom scatter factor and output factor values for MLC were different that those for jaws. Output factor data for jaw-mMLC combination for 6 MV photons is given in Table 4. The output factor values for MLC were different that those for jaws.

Table 1. Output factor comparison between jaw and MLC for 6 MV photons

Field size [cm×cm]	Output factors 6 MV photons (millennium 120 MLC)					
	Sc		Sp		OPF	
	Jaw	MLC	Jaw	MLC	Jaw	MLC
3×3	0.911	0.880	0.973	1.023	0.887	0.900
4×4	0.941	0.921	0.982	1.008	0.924	0.928
5×5	0.954	0.941	0.986	1.005	0.941	0.945
6×6	0.967	0.956	0.988	1.003	0.955	0.959
8×8	0.985	0.981	0.995	1.001	0.980	0.981
10×10	1.000	1.000	1.000	1.000	1.000	1.000
15×15	1.023	1.032	1.010	1.000	1.033	1.032
20×20	1.036	1.052	1.018	1.002	1.054	1.054
25×25	1.045	1.067	1.021	1.004	1.068	1.071
30×30	1.054	1.081	1.025	1.002	1.080	1.083
35×35	1.058	1.087	1.026	1.006	1.086	1.093
40×40	1.065	1.088	1.025	1.004	1.092	1.093

Discussion

Bedford et al. [1] described the commissioning of a Pinnacle treatment planning system. Four Elekta linear accelerators were commissioned for external beam photons. Measured data were used to derive parameter values for the Pinnacle beam model by (1) fitting a Monte Carlo model of the accelerator head to measured data and then extracting the parameters for the Pinnacle beam model, and by (2) using the auto-modeling facility within Pinnacle. Both of these methods yielded dose distributions in accord with published recommendations. A separate small-field beam model, customized for an in-house compact blocking system, was also created, which satisfied appropriate acceptance criteria for stereotactically guided conformal brain treatments. Inhomogeneous, oblique, asymmetrical and irregular fields were also assessed, with

calculated and measured doses agreeing to within $\pm 3\%$. Dose–volume histogram calculation was found to be accurate to within $\pm 5\%$ dose or volume for a grid size of $4\text{ mm} \times 4\text{ mm} \times 4\text{ mm}$, with better accuracy being achieved for finer grids. Isocentric doses were compared between Pinnacle’s collapsed cone convolution algorithm and the Bentley–Milan algorithm within the Target-2 treatment planning system. Dose differences were generally less than 3% in the dose prescribed, with larger values for breast plans, where the Pinnacle algorithm calculated scatter more accurately. Pelvic and thoracic plans were also verified using an anthropomorphic phantom, with local dose differences between calculated and delivered dose of up to 8%, but mainly less than 3%, and with no systematic difference. Ionization chamber verifications using START and RT-01 trial procedures demonstrated differences between calculated and measured

Table 2. Output factor comparison between jaw and MLC for 10 MV photons

Field size [cm×cm]	Output factors 10 MV photons (millennium 120 MLC)					
	Sc		Sp		OPF	
	Jaw	MLC	Jaw	MLC	Jaw	MLC
3×3	0.868	0.881	0.993	0.981	0.862	0.865
4×4	0.926	0.928	0.986	0.987	0.913	0.916
5×5	0.946	0.948	0.989	0.989	0.936	0.938
6×6	0.961	0.961	0.990	0.993	0.952	0.954
8×8	0.984	0.984	0.994	0.996	0.978	0.980
10×10	1.000	1.000	1.000	1.000	1.000	1.000
15×15	1.024	1.023	1.010	1.011	1.034	1.034
20×20	1.035	1.036	1.018	1.019	1.053	1.055
25×25	1.044	1.045	1.023	1.025	1.069	1.071
30×30	1.051	1.055	1.028	1.026	1.081	1.083
35×35	1.056	1.059	1.029	1.029	1.086	1.090
40×40	1.061	1.060	1.029	1.028	1.092	1.090

Table 3. Output factor comparison between jaw and MLC for 23 MV photons

Field size [cm×cm]	Output factors 23 MV photons (millennium 120 MLC)					
	Sc		Sp		OPF	
	Jaw	MLC	Jaw	MLC	Jaw	MLC
3×3	0.844	0.916	0.968	0.979	0.817	0.897
4×4	0.901	0.922	0.980	0.986	0.883	0.909
5×5	0.928	0.929	0.986	0.994	0.915	0.924
6×6	0.948	0.948	0.989	0.993	0.937	0.942
8×8	0.977	0.977	0.995	0.996	0.972	0.973
10×10	1.000	1.000	1.000	1.000	1.000	1.000
15×15	1.017	1.024	1.005	1.004	1.022	1.029
20×20	1.035	1.033	1.011	1.009	1.046	1.043
25×25	1.053	1.052	1.019	1.017	1.072	1.070
30×30	1.064	1.062	1.023	1.022	1.088	1.085
35×35	1.073	1.072	1.028	1.025	1.103	1.099
40×40	1.079	1.078	1.031	1.026	1.113	1.107

doses of less than 2%. Following satisfactory performance in the commissioning process, Pinnacle has now been introduced into routine clinical use.

A new Pinnacle 3D treatment-planning system software release has recently become available (v7.4, Philips Radiation Oncology Systems, Milpitas, CA), which supports modeling of rounded multileaf collimator (MLC) leaf ends; it also includes a number of other software enhancements intended to improve the overall dose calculation accuracy. Cadman [2] provided a general discussion of the dose calculation algorithm and new beam-modeling parameters. The accuracy of a diode dosimeter was established for measurement of MLC-shaped beam profiles required by the new software version by comparison with film and ion chamber measurements in various regions of the field. The results suggest that a suitable diode or other small volume dosimeter with

appropriate energy sensitivity should be used to obtain profiles for commissioning the planning system. Film should be used with caution, especially for larger field profile measurements. The dose calculation algorithm and modeling parameters chosen were validated through various test field measurements including a bar pattern, a strip pattern, and a clinical head and neck IMRT field. For the bar and strip patterns, the agreement between Pinnacle calculations and diode measurements was generally very good. These tests were helpful in establishing the new model parameter values, especially tongue and groove width, additional interleaf leakage, rounded leaf tip radius, and MLC transmission. For the clinical head and neck field, the comparison between Pinnacle and film measurements showed regions of approximately 2 cGy under or overdose. However, the Pinnacle calculations agreed with diode measurements at all points to within 1 cGy or 1% of the maximum dose for the field (67 cGy). The greatest discrepancy between film and diode measurements for the clinical field (maximum of 2.8%) occurred in low-dose regions in the central part of the field. The disagreement may

Table 4. Output factor comparison between jaw and micro-MLC for 6 MV photons

Field size [mm×mm]	Output factors 6 MV photons (micro-MLC) OPF	
	Jaw	mMLC
6×6	0.453	0.442
12×12	0.711	0.705
18×18	0.765	0.763
24×24	0.774	0.773
30×30	0.796	0.794
36×36	0.849	0.845
42×42	0.872	0.860
60×60	0.926	0.913
80×80	0.968	0.957
100×100	1.000	1.000

be due to the over-response of film to scattered radiation in the low-dose regions, which have significant shielding by the MLCs.

Pelagade et al. [6] presented their experience in the commissioning and quality assurance (QA) of PrecisePLAN treatment planning system for intensity modulated radiotherapy (IMRT) using multileaf collimator (MLC), step and shoot technique. The data were obtained during the installation, acceptance test procedure, and commissioning of the unit. The dosimetric data were taken using CC13 pin-point ion chamber (Scanditronix Wellhofer). The beam profiles for various field sizes in the x, y, and diagonal directions were measured. The basic dosimetry parameters such as percentage depth dose (PDD), off-axis ratio (OAR), output factor, and scatter factors needed for patient treatment were evaluated. Point dose measurement and fluence maps did the treatment dose verification.

Deshpande et al. [4] commissioned the the Konrad Inverse treatment planning system. The basic beam data required for commissioning the system were generated on Siemens Oncor impression Plus linear accelerator. The quality assurance of relative and absolute dose distribution was carried out before clinical implementation. The salient features of Konrad planning system, like dependence of grid size on dose volume histogram (DVH), number of intensity levels and step size in sequencer were studied quantitatively and qualitatively. To verify whether the planned dose (from treatment planning system – TPS) and delivered dose are the same, the absolute dose at a point was determined using CC01 ion chamber and the axial plane dose distribution was carried out using Kodak EDR2 in conjunction with Omnipro IMRT phantom and OmniPro IMRT software from Scanditronix Wellhofer. To obtain the optimum combination in leaf sequencer module, parameters like number of intensity levels, step size were analyzed. The difference between pixel values of optimum fluence profile and the fluence profile obtained for various combinations of number of intensity levels and step size was compared and plotted. The calculations of the volume of any RT structure in the dose volume histogram were compared using grid sizes 3 mm and 4 mm. The measured and planned dose at a point showed good agreement (3%) except for a few cases wherein the chamber was placed in a relatively high dose gradient region. The axial plane dose distribution using film dosimetry showed excellent agreement (correlation coefficient >0.97) in all the cases. In the leaf sequencer module, the combination of number of intensity level 7 with step size of 3 was the optimal solution for obtaining deliverable segments. The RT structure volume calculation was found to be more accurate with grid

size of 3 mm for clinical use. Thus a study regarding various aspects of commissioning of the Konrad inverse planning system for IMRT was presented.

We investigated the differences in beam parameters like PDD, cross profiles and output factors between jaws and MLC for 6, 10 and 23 MV photon beams. Disagreement in PDD was observed in the build-up region and at higher depths. The beam profiles did not match in the penumbra region. Collimator scatter factors, phantom scatter factors and output factors showed disagreement for both millennium MLC and micro-MLC. This could have implication for intensity modulated radiation therapy and stereotactic radiosurgery in terms of dose in the build up region, exit dose, dose to the planning target volume (PTV) and organ at risk (OAR). Impact of these dosimetric differences between jaw and MLC needs to be further studied in terms of dose volume histograms for PTV and OAR and it's further impact on tumor control probability (TCP) and normal tissue complication probability (NTCP).

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