Estimation of Jaw and MLC Transmission Factor Obtained by the Auto-modeling Process in the Pinnacle3 Treatment Planning System

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Radiation treatment techniques using photon beam such as three-dimensional conformal radiation therapy (3D-CRT) as well as intensity modulated radiotherapy treatment (IMRT) demand accurate dose calculation in order to increase target coverage and spare healthy tissue. Both jaw collimator and multi-leaf collimators (MLCs) for photon beams have been used to achieve such goals. In the Pinnacle3 treatment planning system (TPS), which we are using in our clinics, a set of model parameters like jaw collimator transmission factor (JTF) and MLC transmission factor (MLCTF) are determined from the measured data because it is using a model-based photon dose algorithm. However, model parameters obtained by this auto-modeling process can be different from those by direct measurement, which can have a dosimetric effect on the dose distribution. In this paper we estimated JTF and MLCTF obtained by the auto-modeling process in the Pinnacle3 TPS. At first, we obtained JTF and MLCTF by direct measurement, which were the ratio of the output at the reference depth under the closed jaw collimator (MLCs for MLCTF) to that at the same depth with the field size 10×10 cm^2 in the water phantom. And then JTF and MLCTF were also obtained by auto-modeling process. And we evaluated the dose difference through phantom and patient study in the 3D-CRT plan. For direct measurement, JTF was 0.001966 for 6 MV and 0.002971 for 10 MV, and MLCTF was 0.01657 for 6 MV and 0.01925 for 10 MV. On the other hand, for auto-modeling process, JTF was 0.001983 for 6 MV and 0.010431 for 10 MV, and MLCTF was 0.00188 for 6 MV and 0.00453 for 10 MV. JTF and MLCTF by direct measurement were very different from those by auto-modeling process and even more reasonable considering each beam quality of 6 MV and 10 MV. These different parameters affect the dose in the low-dose region. Since the wrong estimation of JTF and MLCTF can lead some dosimetric error, comparison of direct measurement and auto-modeling of JTF and MLCTF would be helpful during the beam commissioning.

Key Words: Jaw transmission factor, MLC transmission factor, Auto-modeling process

INTRODUCTION

Radiation treatment techniques using photon beam such as three-dimensional conformal radiation therapy (3D-CRT) as well as intensity modulated radiotherapy treatment (IMRT) demand accurate dose calculation in order to increase target coverage and spare healthy tissue. Both jaw collimator and multi-leaf collimators (MLCs) for photon beams have been used to achieve such goals.[1)]

Kehwar TS et al. suggested that the effects of the positions of the MLC leaves play an important role in dosimetry because they affect scatter factors.[2)] And Chow JC et al. found that there is a decrease of dose in the 'edge' region of the protruded leaves' side in the cross-line profile when both the upper and lower portions of leaves are moved out forming a gap for the profile.[3)]

Recently, concern of the dose calculation in the low-dose region has been increased because relatively inaccurately calcu-

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lated dose can lead to toxicity such as radiation pneumonitis in thoracic radiotherapy.[4] Jang et al. reported secondary radiation from MLCs, which is a major cause of low-dose underestimation, contributes a significant portion of low dose in IMRT plans where the region was covered with or bordered on the MLCs.[5] Hence both jaw transmission factor (JTF) and MLC transmission factor (MLCTF) are major parameters in the dose calculation of the low-dose region.

Many commercial treatment planning system (TPS) supporting those field-shaping equipments have each algorithm available to predict the intended dose distribution dosimetrically.[6,7] Dose calculation algorithms can be categorized into three groups: correction-based algorithms, model-based algorithms, and the direct Monte Carlo.[8] The Pinnacle3 (Philips Medical System, Netherland), which is used in our hospital, is a commercial TPS using the convolution/superposition dose calculation algorithm, which is belonged to the model-based algorithms. In the Pinnacle3, the measured beam data is used to characterize the beam attributes, which determine the model parameters like JTF and MLCTF which are iteratively adjusted during the modeling process so that the dose are computed by the model. They can also be directly obtained measuring each factor by the recommended method. The difference of JTF and MLCTF between the two different methods can reflect on the clinical dose difference depending on how much sensitive the commercial TPS is from the point of dosimetrical view.

In this paper we obtain JTF and MLCTF using the two different methods and evaluate the dose difference through phantom and patient study in the 3D-CRT plan, where the tips of MLCs stay longer in the same position than for IMRT.

MATERIALS AND METHODS

1. Jaw transmission factor by direct measurement

According to the Pinnacle3 Physics reference guide, most of the measurements are recommended at 10 cm depth as a reference depth in the water phantom. The JTF recommended in the Pinnacle can be obtained by the ratio of the output at the reference depth under the closed jaw collimator to that at the same depth with the field size 10×10 cm^2 in the water phantom. But the photon beam of the Varian iX (Varian Medical, Palo Alto, CA) cannot be clinically delivered with one pair of the jaw collimators exactly closed due to the safety. So the beam was delivered with the field size 0.4×10 cm^2, which is the minimum field size along the X-direction, when the jaw collimator was rotated by 90-degree (i.e. X-jaw=10 cm and Y-jaw=0.4 cm). Then we used the 'XY jaw transmission equal' option with measuring one JTF. In this measurement, the source to surface distance (SSD) was 100 cm and the center of the field was shifted by Y, which is Y1-0.2 cm. For example, when Y1=10.4 cm and Y2=−10 cm, the center of the field was positioned at 10.2 cm, as shown in the Fig. 1. We measured the output from Y=2.2 cm by 1 cm up to 10.2 cm with the Farmer-type ion chamber (FC65-G, IBA Dosimetry, Germany), whose cavity volume is 0.65 cm3, whose cavity length 23.1 mm, and whose cavity radius 3.1

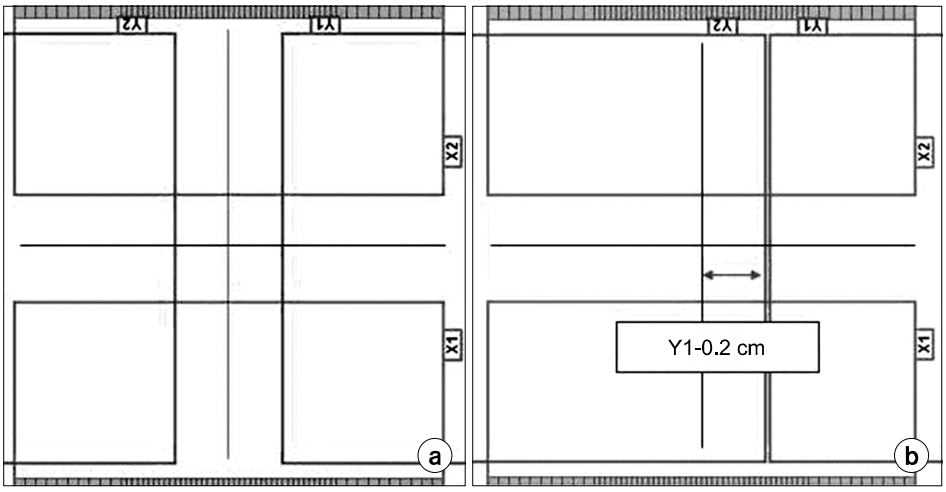

Fig. 1. Geometrical setting for JTF measurement. The intersecting point is the beam center (a) For the output under the open field with the field size 10×10 cm^2 (b) For the output under the closed field with the actual field size 0.4×10 cm^2, where the center of the field is shifted by the Y1-0.2 cm.

mm.

2. MLC transmission factor by direct measurement

MLCTF can be obtained using the similar method to the JTF. With the field size by the jaw collimator 10×10 cm^2 fixed, MLCTF is the ratio of the output at the reference depth under the closed MLC to that at the same depth with the field size 10×10 cm^2 by the MLCs. In this measurement, the same ion chamber was used, and SSD was also 100 cm. Since the Millenium 120 MLCs have the rounded leaf end, the radiation through the rounded leaf end region will be different from that through the center area, when the MLC comes in exact contact with each pair. In order to avoid the leakage due to the geometry of the leaf end, the line where MLCs were in exact contact with each pair was outside the field made by the MLCs. However, since the length of each MLC was 14 cm, the contact line was limited within about 7 cm. According to our measurement, the position of the line, where the output was a minimum, was between 6 cm and 7 cm.

3. JTF and MLCTF by the auto-modeling process

All data required to commission the photon beam of the LINAC, which was a Clinac iX, was put in our commercial TPS, Pinnacle3. Because the Pinnacle3 photon dose algorithm is model-based rather than measurement-based, the software uses the imported measured data only for comparison with the dose profile it calculates for the same measurement geometry. By iteratively adjusting the dose model parameters and evaluating the quality of the match between the measured and computed depth doses and profiles, we can create a dose model which accurately characterizes the output of our machine. The manual options as well as the automatic options can be used during the modeling process in order to obtain more accurate dose calculation via optimizing the parameters. The E_TuneAllInSections sequence was used during the auto-modeling process, which tunes the electron contamination parameters in conjunction with the spectrum. It also optimizes jaw transmission, MLC transmission, and arbitrary fluence profiles.

In order to maximize the dosimetrical difference due to the transmission factor obtained between by direct measurement and by auto-modeling process, the photon beam model in this measurement was generated with only the auto-modeler without a manual process. In the photon beam model by direct measurement all the parameters were exactly the same with those in the photon beam model by auto-modeling process except that JTF and MLCTF by auto-modeling process were replaced by those by direct measurement.

4. Phantom study

In the phantom study, we measured beam profiles and the output of our machine. For the beam profile, the compact ion chamber CC13 (IBA dosimetry, Germany), whose cavity volume was 0.13 cm^3, whose cavity length 0.58 cm, and whose cavity radius 0.3 cm, was used in the Blue Phantom (Scanditronix-Wellhofer, Germany) at the 5 cm depth. The source-to-surface distance (SSD) was 100 cm, the field size made by jaw collimator was 20×20 cm^2, and the field size by MLCs was arbitrarily made using the Shaper (Varian Medical Systems, Palo Alto, CA) as shown in Fig. 2. In order to reduce the effects on the direction of the chamber length, the ion chamber axis was aligned to the direction of MLC. All the beam profiles were normalized at some points where the dose was measured with the Farmer-type ion chamber, FC65-G. The difference of these beam profiles generated from the Pinnacle3 with the two different methods was compared. The planar dose

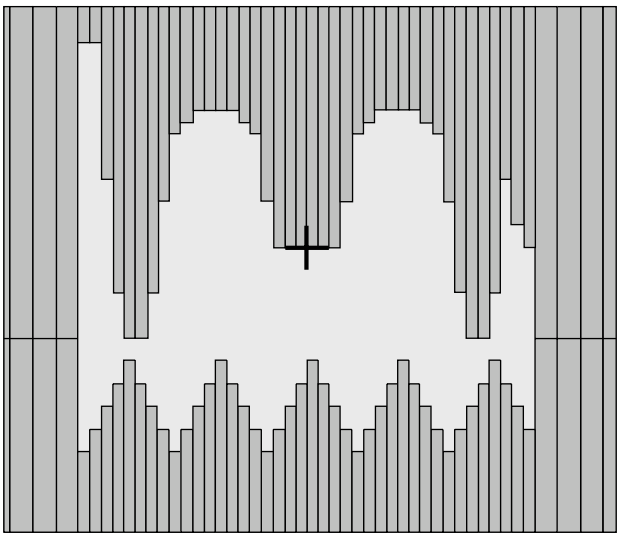

Fig. 2. The field size 20×20 cm^2 by jaw collimators and arbitrarily by MLCs. The cross is the beam center. The line where MLCs were in exact contact with each pair was shifted by 4 cm.

difference was also generated in the Pinnacle3, where SSD is equal to 100 cm, and the source-to-detector distance (SDD) is 105 cm.

5. Patient study

One thoracic CT image set without lung cancer was selected in our clinic. Three virtual organs were contoured arbitrarily as shown in Fig. 3. The treatment field was used with 10 MV of 2 ports, anterior-posterior (AP) and posterior-anterior (PA). MLC files of each port were copied using the Shaper, from which MLC field was put in our machine. The field size by jaw collimator was manually generated to 10.1×10.4 cm^2. The field size was much larger than the size of the virtual planning tumor volume (PTV) plus the margin considering the penumbra in order to evaluate the effect of the MLCTF on the dose calculation. A region of interest (ROI) as a gross tumor volume (GTV) was contoured, which was a vir_Lung_GTV ROI. Another ROI as a PTV was expanded by 1 cm from the vir_Lung_GTV, which was a vir_Lung_PTV. And the last ROI (= a vir_Lung_Ring) was created as a ring expanded by 1 cm from the PTV, which were the region covered by mainly MLCs. The dose distribution and dose volume histograms (DVHs) were also compared under the same monitor units (MUs). The volumes of each ROIs are 49.4697 cm^3, 178.066 cm^3, and 222.617 cm^3.

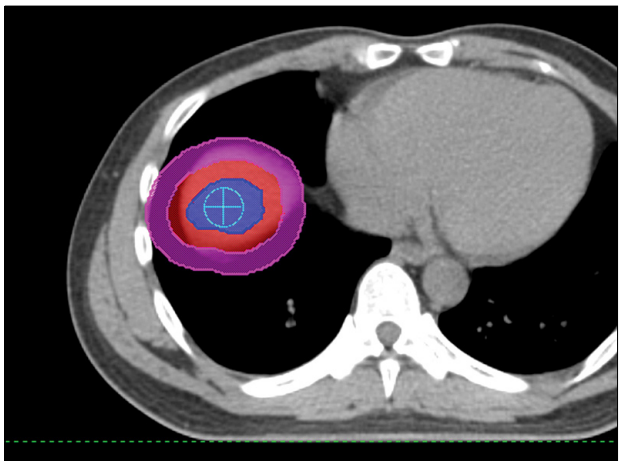

Fig. 3. An axial CT image of the patient. The cross denotes the isocenter of the beam. The blue does a virtual GTV (a vir_Lung_GTV), the red a virtual PTV (a vir_Lung_PTV), and the pink a virtual ROI (a vir_Lung_Ring).

RESULTS AND DISCUSSION

1. JTF and MLCTF by direct measurement

The behavior of the JTF of 6 MV and 10 MV by direct measurement was shown in Fig. 4. We used the trial function F(x) to obtain the asymptotic JTF as x goes to infinite.

$$F(x) = A \cdot X^{-B} - C,$$

where X means the distance from the isocenter to the beam center, F(X) means the log of the JTF depending on the position of the beam center, and A, B, and C are the fitting parameters.

In our measurement, for 6 MV [10 MV], A=2.639 [2.262], B=0.5911 [0.8721], and C=6.232 [5.819]. Hence the JTF of 6 MV was 0.0019831, which was very similar to that obtained by the auto-modeling process, which was 0.001966. However, except the JTF of 6 MV, the other parameters were quite different as shown in Table 1. The JTF of 10 MV by direct measurement was 0.002971, while that by auto-modeling process was 0.0104310. While the ratio of the JTF of 6 MV to that of 10 MV by auto-modeling was 0.190, the ratio of the JTF of 6 MV to that of 10 MV by direct measurement was 0.662, which was more reasonable considering the mass attenuation coefficient depending on the different energy.[9)]

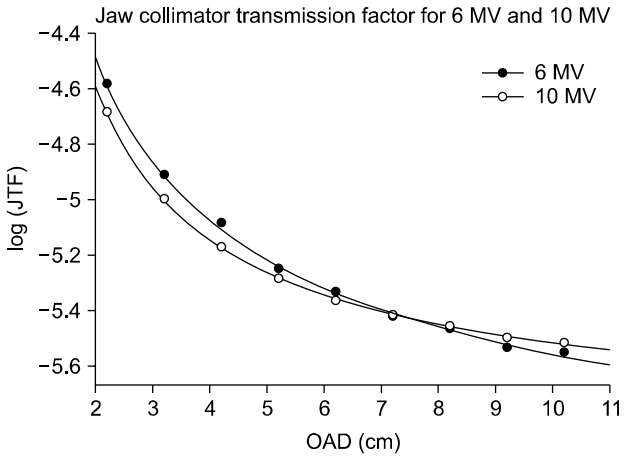

Fig. 4. The behavior of JTF of 6 MV and 10 MV by direct measurement. The white denotes JTF of 10 MV, and the black one does that of 6 MV.

Table 1. JTF and MLCTF comparison.

	Energy	Auto-modeling	Direct measurement	Difference*
JTF	6 MV	0.001983	0.001966	-1.70×10^{-5}
	10 MV	0.010431	0.002971	-7.46×10^{-3}
MLCTF	6 MV	0.00188	0.01657	1.47×10^{-2}
	10 MV	0.00453	0.01925	1.47×10^{-2}

*Difference=value by direct measurement - value by auto-modeling.

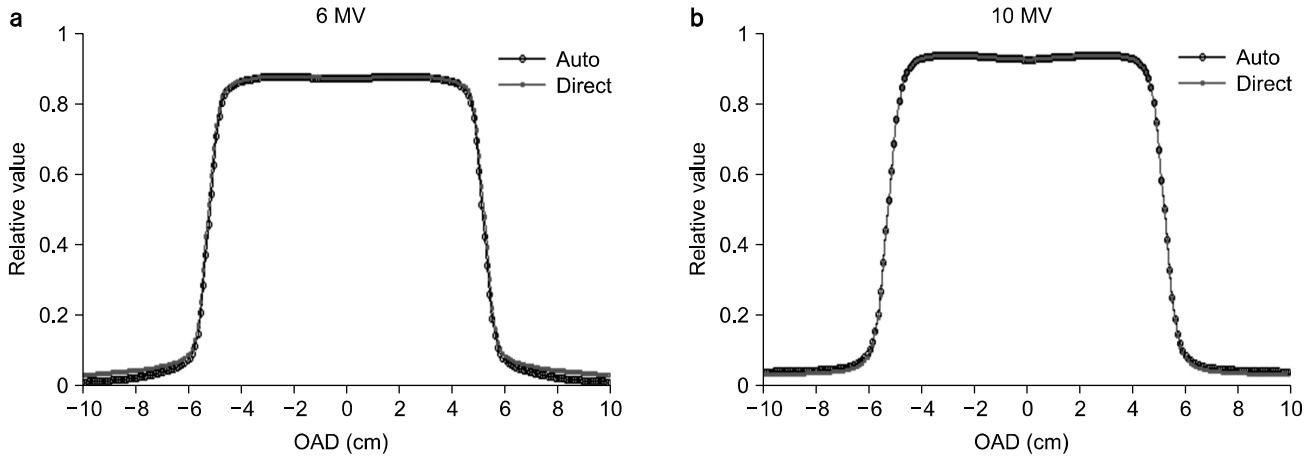

Fig. 5. Beam profile at the 5 cm depth with field size of both Jaw and MLC 10×10 cm^2. (a) 6 MV (b) 10 MV.

On the other hand, the MLCTF of 6 MV by direct measurement was 0.01657, while that by auto-modeling was 0.00188. And those of 10 MV were 0.1925 and 0.00453, respectively. Even though the ratio of the MLCTF by direct measurement was not significantly different from that by auto-modeling process, the MLCTFs by auto-modeling process were unreasonable considering the material of the jaw collimator and the MLCs.

Hence we found that JTF and MLCTF of photon beams by direct measurement were more reasonable than those by auto-modeling process and this huge discrepancy was probably due to insufficient iteration during the auto-modeling process.

2. Phantom study

The beam profile by direct measurement (with CC13 ion chamber at 5 cm depth with field size of both Jaw and MLC 10×10 cm^2) and by auto-modeling process of 6 MV and 10 MV was shown in Fig. 5. The dose difference is dominant around the penumbra region and the out-of-field. The total transmission factor of 6 MV equal to JTF multiplied by MLCTF is larger than that of 10 MV, so the dose difference of 6 MV has larger than that of 10 MV in the out-of-field.

The planar dose difference per unit dose with the field shape as shown in Fig. 2 was shown in Fig. 6 with 2 mm resolution. The most different region was around the tip of each MLC.[3] The dose difference of 6 MV was larger than that of 10 MV, and which were about 4% for 6 MV, and 2～3% for 10 MV. For the region covered by only MLCs the dose differences were 1.5～2% for 6 MV, and around 1.5% for 10 MV. For the fully open region the dose difference is negligibly below 0.5%. This discrepancy was because the MLCTF difference of 6 MV was larger than that of 10 MV. The ratio of MLCTF of 6 MV by direct measurement to that by auto-modeling process was about 8.8, while the ratio for 10 MV was about 4.25. The dose difference in the open region was also negligible because there was no different parameter between the two methods.

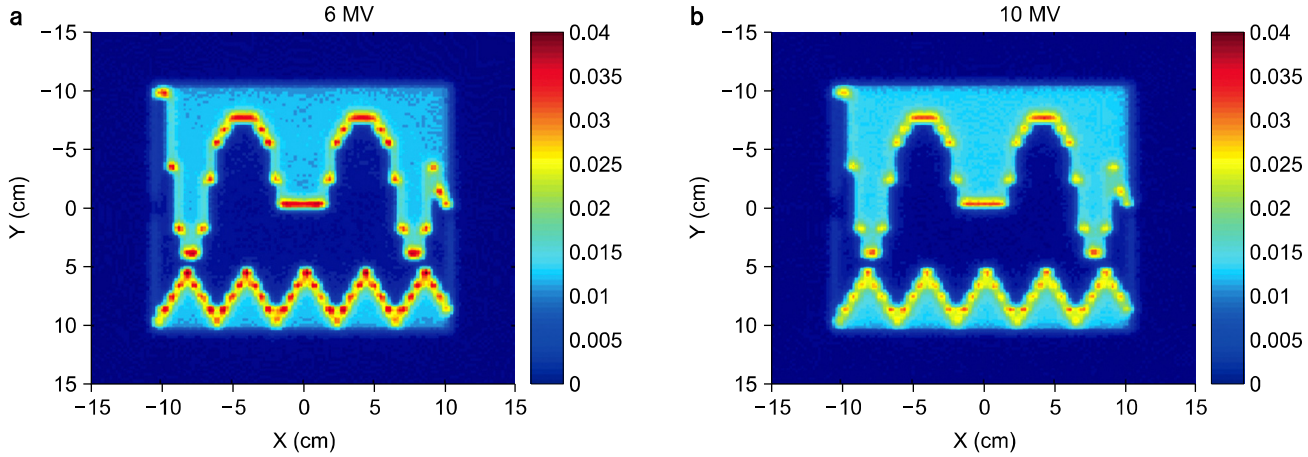

Fig. 6. Planar dose difference per unit dose with the field shape as shown in Fig. 2. (a) 6 MV (b) 10 MV.

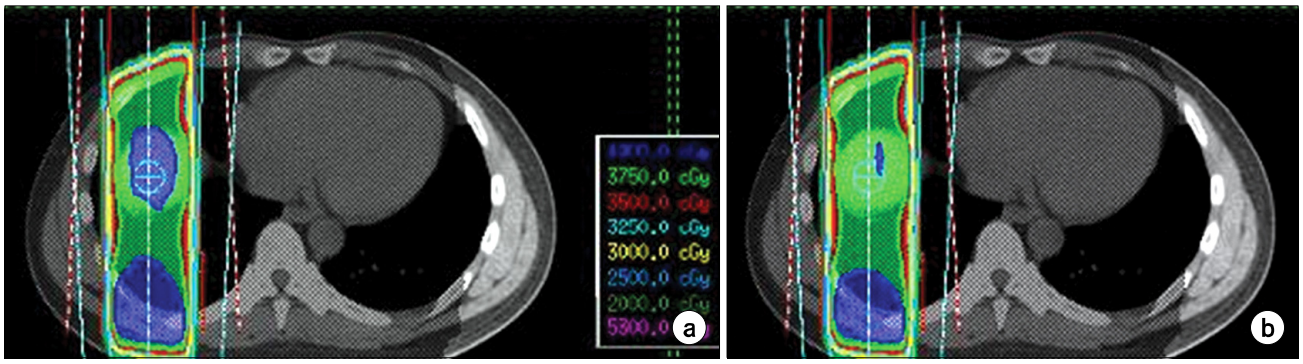

Fig. 7. The comparison of the dose distribution between (a) by the direct measurement and (b) by the auto-modeling process under the same MUs (105 MUs for AP port and 115 MUs for PA one).

3. Patient study

The result of patient study was almost same with that of phantom study. The dose distribution between by the direct measurement and by the auto-modeling process was compared in Fig. 7. The dose difference around the target was relatively small. Under the same 220 MUs, the point dose at the iso-center for the direct measurement was 200.58 cGy per fraction while that at the same spot for the auto-modeling process was 200.36 cGy per fraction. Hence the dose difference at the iso-center was only 0.22 cGy per fraction which was within 0.11%. However the dose difference in the ring ROI was relatively large as shown in the Fig. 8. The mean dose in the vir_Lung_Ring for the direct measurement was 112.36 cGy per fraction and that for the auto-modeling process 109.34 cGy

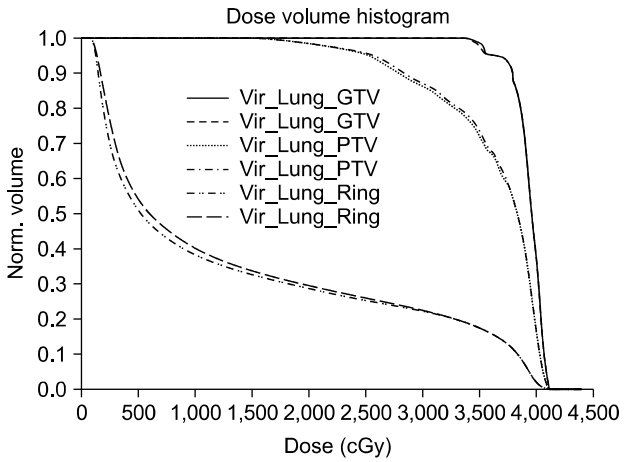

Fig. 8. The results of the DVHs for the case of the patient study.

per fraction. Hence the mean dose difference in the ring ROI was 3.02 cGy per fraction which was about 2.69%. Especially at a certain point around the boundary of the ring the point dose for the direct measurement was 52.74 cGy per fraction while that at the same point for the auto-modeling process was 48.00 cGy. And so the dose difference was 4.27 cGy per fraction and the percentage difference was over 8%. The lower dose region has more significant different dose value between the two different methods.

CONCLUSION

In conclusion, comparison of direct measurement and auto-modeling of JTF and MLCTF would be helpful during the beam commissioning because the auto-modeler without manually controlling the beam model can generate the unreasonable JTF and MLCTF values.

Even though we evaluated the dosimetrical effect of the beam parameter, JTF and MLCTF, in 3D-CRT, the dose difference for IMRT can be more severe than that for 3D-CRT. One of the reasons is that IMRT needs more MUs. Another is that the boundaries between PTV and normal organ have steep dose gradient due to many repeatedly open-and-closed status of MLCs. However, the point dose difference can be on the contrary from the point of local point dose view because the tips of MLCs stay for a relatively long time during the beam-on time. Therefore for 3D-CRT as well as for IMRT accurate JTF and MLCTF are very important and that is why commissioning should be carefully done.

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피나클치료계획시스템에서 자동모델화과정으로 얻은 Jaw와 다엽콜리메이터의 투과 계수 평가

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세기조절방사선치료(IMRT)뿐만 아니라 3차원 입체조형치료(3D-CRT)와 같이 광자선을 이용한 방사선 치료 기술은 방사선을 받아야 하는 표적의 면적을 충분히 증가시키면서, 동시에 정상 조직은 방사선으로부터 보호하기 위하여 정확한 선량 계산을 필요로 한다. Jaw 콜리메이터와 다엽 콜리메이터가 그러한 목적을 위해서 사용되어 왔다. 우리 기관에서 사용하는 피나클 치료계획시스템은 모델기반의 광자선량 알고리듬을 사용하기 때문에 Jaw 콜리메이터 투과계수(JTF)와 다엽 콜리메이터 투과계수(MLCTF)와 같은 모델변수들의 집합이 측정된 데이터로부터 결정된다. 그러나, 이러한 자동모델화과정에 의해서 얻어진 모델변수들이 직접 측정하여 얻은 것들과 다를 수 있는데, 이는 선량분포에 영향을 줄 수 있다. 그래서, 이 연구에서 우리는 피나클 치료계획시스템에서 자동모델화과정에 의해 얻은 JTF와 MLCTF를 평가하였다. 먼저 우리는 이 연구에서 Jaw 콜리메이터 투과계수(JTF)와 다엽 콜리메이터 투과계수(MLCTF)를 직접 측정하여 얻었는데, 이것은 물팬톰 내 기준깊이에서 조사면이 0×0 cm^2일 때의 선량과 10×10 cm^2일 때의 선량의 비로 얻었다. 또한, JTF와 MLCTF는 치료계획시스템내 자동모델화 과정에 의해서도 얻어서, 이 값들이 3차원 입체조형치료시에 선량에 어떠한 영향을 끼치는지 팬톰 연구와 환자 연구를 통해서 평가하였다. 직접 측정한 경우 JTF는 6 MV의 경우에 0.001966, 10 MV의 경우에는 0.002971이었고, MLCTF는 6 MV의 경우에 0.01657, 10 MV의 경우에 0.01925이었다. 한편, 자동모델화 과정에 의해 얻은 경우, JTF는 6 MV의 경우에 0.001983, 10 MV의 경우에는 0.010431이었고, MLCTF는 6 MV의 경우에 0.00188, 10 MV의 경우에 0.00453이었다. JTF와 MLCTF의 경우에 직접 측정한 것은 자동모델화 과정에 의해 얻은 값과 큰 차이를 보였으나, 6 MV와 10 MV의 선질을 고려하면, 보다 합리적이었고, 이러한 값의 차이는 낮은 선량의 영역에서 선량에 영향을 미쳤다. JTF와 MLCTF의 잘못된 값은 선량의 오차를 다소 발생시킬 수도 있기 때문에, JTF와 MLCTF를 자동모델화 과정에 의해서 얻은 값과 직접 측정하여 얻은 값을 비교하는 것은 빔커미셔닝 단계에서 도움이 될 것이다.

중심단어: Jaw 콜리메이터 투과계수(JTF), 다엽 콜리메이터 투과계수(MLCTF), 자동모델화과정(Auto-modeling process)