

## Impact of dose calculation algorithm on radiation therapy

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### Abstract

The quality of radiation therapy depends on the ability to maximize the tumor control probability while minimize the normal tissue complication probability. Both of these two quantities are directly related to the accuracy of dose distributions calculated by treatment planning systems. The commonly used dose calculation algorithms in the treatment planning systems are reviewed in this work. The accuracy comparisons among these algorithms are illustrated by summarizing the highly cited research papers on this topic. Further, the correlation between the algorithms and tumor control probability/normal tissue complication probability values are manifested by several recent studies from different groups. All the cases demonstrate that dose calculation algorithms play a vital role in radiation therapy.

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**Key words:** Dose calculation; Algorithm; Radiation therapy; Tumor control probability; Normal tissue complication probability

**Core tip:** This paper is a review of the impact of current

commercial dose calculation algorithms on radiation therapy, with a focus on discussing the impact on tumor control probability and normal tissue complication probability.

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### INTRODUCTION

The quality of radiation therapy depends on the ability to maximize the tumor control probability (TCP) while minimize the normal tissue complication probability (NTCP) at the same time. Since these two quantities are directly dependent on the absorbed dose in the targets and in the organs at risk (OARs) respectively, accurate knowledge of dose distribution within the patient are crucial in radiation therapy. International Commission on Radiation Units and Measurements (ICRU)<sup>[1]</sup> has recommended an overall dose accuracy within 5%. Considering the uncertainties resulting from patient setup, machine calibration and dose calculation from treatment planning systems, it is necessary to have a dose calculation algorithm that can predict dose distribution within 3% accuracy.

Accurate calculation of dose distribution in an inhomogeneous medium such as human body is a complicated task, especially for tumors located in the lung. To date, only the Monte Carlo method is considered to be the most accurate algorithm for dose calculation but it requires the greatest processing time. Apart from Monte Carlo method, all other methods make different degrees of approximation and simplification which lead to much faster calculation speed but also result in less accurate dose distribution comparing with the Monte Carlo simulation.

The purpose of this study is to review the effect of dose calculation algorithms on the radiation therapy for different disease sites and special focus is given for the lung region. As mentioned in the American Association of Physicists in Medicine (AAPM) Report No. 85<sup>[2]</sup>, the level of dose differences can be detected clinically. In order to quantify the clinical effects, we review the studies on the correlation of dose calculation algorithms with computed values of tumor control probability and normal tissue complication probability. The impact of the accuracy of the algorithms is directly related to the quality of radiation therapy.

## DOSE CALCULATION ALGORITHMS

The Monte Carlo dose calculation method is considered to be the most accurate algorithm and has always been used as the generation of benchmark dose distribution with which to compare the results of other less-computer-intensive dose calculation methods<sup>[3]</sup>. The Monte Carlo method uses photon and electron transport physics to consider the trajectories of individual particles and thus the pattern of dose deposition. Each particle's history is determined by the random number generator and millions of particles' histories are traced. The dose distribution is built by summing the energy deposition in each particle's history.

Apart from the Monte Carlo simulation, all other commonly used dose calculation algorithms can be categorized into two groups<sup>[2,4,5]</sup>: (1) Methods based on equivalent path length (EPL)<sup>[6]</sup> scaling or equivalent tissue-air ratio (ETAR)<sup>[7]</sup> for inhomogeneity corrections. In these methods the changes in lateral transport of electrons are not modeled; and (2) Methods based on convolution techniques, in which the inhomogeneities are handled either by an equivalent path length correction or scaled kernels and the lateral electron transport is considered in an approximate way. In this work, these two types of algorithms are referred to as type (1) and type (2) methods. In type (1) methods, the equivalent path length correction is a one-dimensional method that takes into account of electron density information along a ray path from the source to the point in question. There are two methods: ratio of tissue-air ratio (RTAR) method<sup>[2]</sup> and power law method which is also referred as modified Batho method<sup>[8]</sup>. These methods correctly account for the change in the attenuation of the primary dose but not in the scatter contribution, thus result in an overestimation of dose when the electron density is less than unity and an underestimation when the electron density is greater than unity. The equivalent tissue-air ratio method is a three-dimensional correction method which is based on full three-dimensional density information acquired from CT images. This method applies a ray trace to determine the change in the primary dose and calculate the scatter dose based on the three-dimensional density data. Although methods in type (1) do not perform an accurate dose distribution calculation in patients, they are still used by

some treatment planning systems for a quick dose calculation to give the planner a rough idea about the absorbed dose and by some dose verification systems to perform a second independent check to catch the gross errors.

In type (2) methods, the model-based convolution/superposition algorithms<sup>[9-13]</sup> are widely used in commercial radiotherapy treatment planning systems (TPSs), which perform dose calculations with accuracies close to the results of Monte Carlo simulation while take much less time. All convolution algorithms have two essential components: one representing the energy imparted to the medium by the interactions of primary photons, called Terma (total energy released per unit mass) and one representing the energy deposited about a primary photon interaction site, the kernel. The kernel can be further separated into two parts: the primary kernel which calculates the primary dose and the scatter kernel which calculates the first and multiple scatter doses. The dose at any point can be calculated from the convolution of the Terma with the kernel. In order to account for tissue heterogeneities in a patient, kernel is scaled by radiological distances which are calculated from the material densities defined by CT images. Rigorously speaking, when the scaled kernel is used, the process is not a convolution any more since the kernel is not invariant in space and it is in fact a superposition of varying kernels with the Terma. The treatment planning systems that use the superposition algorithm include, for example, XiO (Elekta, Inc.). Several variations of the convolution/superposition algorithms exist today and two typical and mostly used ones are collapsed cone convolution (CCC) and pencil beam convolution (PBC) techniques<sup>[4]</sup>. The collapsed cone convolution method uses a polyenergetic Terma and kernel, where the kernel is represented analytically and expressed in polar coordinates. There are a finite number of polar angles with respect to the primary beam. The interaction site can be considered to be at the apex of a set of radially directed lines spreading out in three dimensions. Each line is considered to be the axis of a cone. The kernel along each line is actually the energy deposited within the entire cone collapsed onto the line. The advantage of the CCC method over standard convolution is that the computation time increase with  $MN^3$  as opposed to  $N^6$ , where  $M$  is the number of cones and  $N$  is the number of voxels along one side of the calculation volume. The treatment planning systems that use the CCC method include, for example, Pinnacle (Philips, Inc.) and Oncentra MasterPlan (Nucletron, Inc.). In the pencil beam convolution method, the dose deposited at a point is calculated as a convolution of Terma with a pencil-shape-like kernel which is derived from the measured beam data. The pencil-beam kernel describes the dose distribution of a very narrow beam entering a water phantom along the beam's central axis. Inhomogeneity correction is performed with an equivalent path length correction for the primary dose contribution and a one-dimensional convolution along fan lines for scattered radiation<sup>[14,15]</sup>. The anisotropic analytical algorithm (AAA)<sup>[16,17]</sup> used by Eclipse TPS

(Varian Medical Systems) is based on the pencil beam convolution technique. The AAA uses spatially variant convolution scatter kernels which are derived from Monte Carlo simulation, and separate modeling for primary photons, scattered photons, and contaminant electrons. Inhomogeneity is handled with radiological scaling of the dose deposition functions in the beamlet direction and electron-density-based scaling of the photon scatter kernels in 16 lateral directions. The final doses are obtained by superposing the doses from the photon and electron convolutions<sup>[18,19]</sup>. The anisotropic analytical algorithm is an attractive option for routine clinical use because of its relatively short computation time and accuracy comparing with the Monte Carlo method.

## COMPARISON OF DOSE CALCULATION ALGORITHMS AND THEIR CLINICAL IMPACT

Comparisons of dose calculation algorithms for clinical treatment disease sites have been studied in many references<sup>[4,19-21]</sup>. In this review, we first summarize the comparisons of dose calculation algorithms for four commonly treated disease sites, which demonstrate that dose calculation algorithms that can calculate dose accurately in inhomogeneous environment are essential for lung tumor treatment. Then we focus on the dose calculation algorithms for lung tumor treatment planning. Different treatment techniques are discussed. Finally we show the correlation of the algorithms with TCP/NTCP.

In Knöös *et al.*<sup>[4]</sup>'s paper, the authors studied the performance of different dose calculation algorithms from five commercial radiotherapy treatment planning systems for four common treatment disease sites: prostate, head and neck, breast and lung. The Monte Carlo algorithm was used as a benchmark for comparison between different algorithms. Increasing the complexity from the relatively homogeneous pelvic region to the very inhomogeneous lung region resulted in less accurate dose distributions. Improvements in the accuracy of dose calculation were observed when the methods taking into account of volume scatter and changes in electron transport were used, that is, when type (2) algorithms were used. That was especially important when the extension of the irradiated volume was limited such as in the breast case and when low densities were presented such as in the lung case. In the prostate case, no significant differences were found in the results calculated with different algorithms. For instance, when 6 MV was used, the dose to 95% of the PTV was in a range of 96.2% to 100.3% for all studied systems, with an average value of 98.2%. Qualitatively, all the plans which were calculated with different methods, were very similar. The similar situation existed in the head and neck case. The average dose per monitor unit (MU) to the PTV was decreased by 1% for the low energy if more accurate methods, *i.e.*, type (2) methods, were used. This difference was not presented

for the higher energy, due to less scatter in the high energy beam. The dose to 95% of the PTV showed no significant change when moving from type (1) methods to type (2) methods for both low and high energies. The ETAR method of type (1) resulted in doses closer to that calculated with type (2) methods, due to the improved scatter integration which took into account the 3D extension of the volume more accurately. In the breast case, two equally weighted opposed tangential beams were used. The average PTV doses were decreased by 0.7% and 1.6% for low and high energies, respectively, when comparing type (1) with type (2) methods. In general, larger differences in dose calculation were found in high energy treatment due to the longer range of electrons, especially in the low density lung tissues. In the pulmonary case, for 6 MV, the average dose per MU to the PTV was decreased by 2.5% when the type (2) methods were used, compared with that calculated with type (1) methods. Changing the energy to high energies increased the difference to 3.7%. The high dose volume within the PTV was decreased by 3.4% and 4.6%, moving from type (1) methods to type (2) methods for low and high energies, respectively. This implies that accurate tumor doses are different from the doses predicted with those methods, and accurate tumor doses needs to be predicted with advanced dose calculation algorithms, *i.e.*, Monte Carlo algorithm. Thus the algorithm directly affects the local control of tumors in lung cancer. That is, less coverage for tumor is presented when more realistic and accurate methods is used. This paper and other references<sup>[19-22]</sup> showed that the dose calculation algorithms have a significant impact on radiation therapy for lung cancer treatment.

Remarkable impact of dose calculation algorithms on radiation therapy has been observed in the treatment of lung cancer, when tissue density correction was taken into account. Differences between dose calculations with and without density corrections in the thoracic region have been reported<sup>[23-28]</sup>. In Xiao *et al.*<sup>[27]</sup>'s paper, a retrospective dosimetric study was carried out based on the treatment plans submitted to Radiation Therapy Oncology Group (RTOG) 0236 clinical trials of non-small-cell lung cancer (NSCLC) treatment with stereotactic body radiotherapy (SBRT). The protocol required each institution to submit two plans: one plan without heterogeneity correction and one plan with heterogeneity correction, with identical MUs. In Xiao *et al.*<sup>[27]</sup>'s study, the authors found that the planning target volume receiving greater than 60 Gy was decreased, on average, by 10.1% when heterogeneity corrections were applied. The maximal dose to any point greater than 2 cm away from the planning target volume increased from 35.2 Gy to 38.5 Gy.

The impact of heterogeneity corrections of dose algorithms on target coverage in the SBRT lung treatment was studied in more details in Ding *et al.*<sup>[22]</sup>'s paper. The dose calculations using four different algorithms were compared with experimental measurements. The pencil beam algorithm with no heterogeneity corrections (PBN-C) and with modified Batho heterogeneity corrections

**Table 1** Calculated percent mean tumor control probability values (ranges in parentheses) for all algorithms as a function of planning target volume volume

PTV bins (cm <sup>3</sup> )	Mean PTV volume (range, cm <sup>3</sup> )	n	EPL-1D	EPL-3D	AAA	CCC	Acuros	MC
4 ≤ v < 10	7.8 (4.8-9.9)	15	100.0 (100-100)	99.9 (99.6-100)	93.1 (76.3-99.8)	91.3 (63.0-99.9)	91.8 (60.8-99.8)	90.5 (51.1-99.9)
10 ≤ v < 20	15.0 (10.4-19.8)	27	100.0 (99.8-100)	99.9 (99.5-100)	91.3 (61.7-100)	91.3 (50.4-100)	91.4 (65.4-99.9)	91.1 (53.2-100)
20 ≤ v < 30	24.3 (20.4-29.6)	29	98.5 (99.8-100)	98.9 (77.6-100)	92.7 (74.9-99.9)	90.5 (46.4-99.9)	90.9 (65.1-99.9)	91.1 (48.4-99.9)
30 ≤ v < 40	34.9 (30.2-39.8)	18	99.8 (97.4-100)	99.6 (98.2-100)	92.0 (63.4-99.9)	92.1 (69.7-99.9)	90.9 (61.6-99.8)	92.4 (56.3-99.9)
40 ≤ v < 60	47.3 (40.2-58.4)	17	99.5 (93.1-100)	99.1 (95.6-100)	92.6 (78.6-99.9)	91.4 (64.4-99.9)	93.6 (77.6-99.9)	92.3 (63.6-99.9)
60 ≤ v < 100	78.0 (60.4-95.9)	16	99.5 (95.6-100)	99.0 (95.8-100)	92.7 (70.7-99.8)	92.8 (66.2-99.9)	93.4 (70.4-99.8)	94.7 (74.6-99.9)
V ≥ 100	162.4 (100.5-360.2)	11	99.2 (96.1-99.9)	98.7 (95.0-100)	96.3 (89.9-100)	95.6 (91.6-99.8)	95.3 (83.0-99.9)	97.1 (88.8-99.9)

PTV: Planning target volume; EPL-1D: 1-D equivalent path-length (pencil beam-type); EPL-3D: 3-D equivalent-path-length (pencil beam-type); AAA: Anisotropic analytical algorithm; CCC: Collapsed cone convolution-superposition; Acuros: Acuros AXB; MC: Monte Carlo. (Cited from Chetty *et al*<sup>[30]</sup> 2013).

(PB-MB), the anisotropic analytical algorithm (AAA) and Monte Carlo simulation were investigated in ten patients' treatment planning. The plans included 8-10 non-opposed photon beams and 2-4 of the beams were non-coplanar. The field sizes ranged from 3.5 cm × 3.5 cm to 6 cm × 6 cm with the mean value close to 4 cm × 4 cm. The mixed 6 and 10 MV energies were used. The authors found that the differences in calculated doses to 95% or 99% of the PTV, between calculations using the PB-NC and the AAA, were within 10% of prescribed dose. Compared to that calculated with the AAA, the minimum doses to 95% of PTV calculated using the PB-MB were overestimated by up to 40% of the prescribed dose. The calculated maximum doses were underestimated by up to 27% using the PB-NC and overestimated by 19% using the PB-MB. The dose distributions near the interface calculated with the AAA agreed with those from Monte Carlo calculations and the measurements.

The above publications demonstrated the impact of dose calculation algorithms on the lung cancer treatment. These comparisons were mainly between type (1) and type (2) methods. The direct comparisons between type (2) algorithms and Monte Carlo simulation have been done extensively. For instance, in Vanderstraeten *et al*<sup>[20]</sup>'s study, the authors compared the accuracy between Monte Carlo, convolution/superposition, and pencil beam dose calculations for intensity modulated radiation therapy (IMRT) of lung cancer, and they found that the convolution/superposition methods showed an excellent agreement with Monte Carlo method for dose calculation within the target structures, whereas the best agreement in OAR doses was found between collapsed cone convolution model and Monte Carlo simulation. Results from pencil beam algorithm were unsatisfying for both target and OARs. In Li *et al*<sup>[28]</sup>'s paper, the authors compared superposition algorithm with Monte Carlo method for SBRT non-small-cell lung cancer treatment and they found that the important dosimetric parameter R50 (ratio of 50% prescription isodose volume to PTV) recommended by RTOG 0813 protocol had 12% difference on average between superposition and Monte Carlo calculations.

All these research studies have demonstrated that

for dose calculation in lung region the advanced type (2) methods are necessary, and the collapsed cone convolution algorithm and anisotropic analytical algorithm are appropriate options for their relative accurate calculation results compared with the Monte Carlo method.

In the above, we have discussed that the different dose calculation algorithms could give different levels of dose distribution accuracy. Further we will discuss that this different levels of accuracy could be detected clinically, which affect the quality of radiotherapy. The American Association of Physicists in Medicine (AAPM) Report No. 85<sup>[2]</sup> on tissue inhomogeneity corrections mentioned that a 5% change in dose may result in a significant change in tumor control probability (TCP) and normal tissue complication probabilities (NTCP). In this report, the authors mentioned two examples<sup>[29]</sup>: A 7% difference in dose delivered to different groups of patients was discovered by a radiation oncologist; and two experiences from the Institut Gustave Roussy, which were related to tumor regression and normal tissue reactions, respectively.

Although it is still a relative new topic, the correlation between dose algorithms and local control, TCP and NTCP, has already been investigated by several groups and more research is expected to be done in the future. In Chetty *et al*<sup>[30]</sup>'s study, 133 NSCLC patients with stereotactic ablative radiotherapy (SABR)-based treatment were chosen for the correlation study. The correction-based pencil-beam algorithm, model-based convolution/superposition algorithm, and Monte Carlo algorithm were applied for dose calculation. TCP was computed using the Marsden model<sup>[31,32]</sup> and associations between dose and outcome were inferred. The authors found that model-based mean TCP's were approximately 8%-9%, 6%-8%, and 3%-5% lower than those of correction-based algorithms for volumes < 60, 60-100, and > 100 cm<sup>3</sup>, respectively, when the same treatment arrangement was applied. This was because that the advanced type (2) methods simulated the dose deposition physics in a more realistic way than that type (1) methods. Further, the maximum decrement in Monte Carlo-based TCP was about 50% for volumes < 30 cm<sup>3</sup>. Variation in TCP ranges among model-based algorithms is due to the differences in the



**Table 2** Relative differences calculated as (without-with)/with density corrections using each algorithm

	Eclipse AAA	OTP CC	Pinnacle CC	XiO Sup	OTP PB	XiO FFT
Combined lungs						
NTCP <sub>Burman</sub>	-0.29	-0.2	-0.22	-0.25	-0.36	-0.45
NTCP <sub>Seppenwoolde</sub>	-0.19	-0.13	-0.12	-0.15	-0.23	-0.3
Mean dose	-0.08	-0.05	-0.05	-0.06	-0.09	-0.13
V <sub>20</sub>	-0.06	-0.06	-0.04	-0.03	-0.05	-0.07
Heart						
NTCP	-0.19	-0.15	-0.15	-0.13	-0.17	-0.21
Mean dose	-0.06	-0.05	-0.05	-0.05	-0.06	-0.09
V <sub>50</sub>	-0.11	-0.1	-0.1	-0.08	-0.08	-0.13
PTV						
Mean dose	-0.06	-0.05	-0.05	-0.05	-0.13	-0.1
D <sub>01</sub>	-0.05	-0.05	-0.04	-0.04	-0.1	-0.11
D <sub>99</sub>	-0.07	-0.04	-0.05	-0.05	-0.14	-0.09
GTV						
Mean dose	-0.07	-0.06	-0.06	-0.06	-0.08	-0.1
D <sub>01</sub>	-0.07	-0.07	-0.06	-0.06	-0.09	-0.11
D <sub>99</sub>	-0.07	-0.06	-0.06	-0.06	-0.07	-0.1

Negative results indicate lower values when no density corrections are included. Eclipse AAA: Eclipse Anisotropic Analytical Algorithm; OTP CC: Oncentra MasterPlan Collapsed Cone algorithm; Pinnacle CC: Pinnacle Collapsed Cone algorithm; XiO Sup: XiO Multigrid Superposition algorithm; OTP PB: Oncentra MasterPlan Pencil Beam algorithm; XiO FFT: XiO Fast Fourier Transform Convolution algorithm. (Cited from Nielsen *et al*<sup>[34]</sup> 2011).

**Table 3** Clinical impact of dose calculation algorithms

Ref.	Tumor site/technique	Algorithms studied	Results/conclusion
Nielsen <i>et al</i> <sup>[34]</sup> , 2011	NSCLC	Eclipse AAA OTP CC Pinnacle CC XiO Sup OTP PB XiO FFT	Differences in dose to target predicted by the different algorithms are of a magnitude. Calculated NTCP values for pneumonitis are more sensitive to the choice of algorithm than mean lung dose and V20
Chandrasekaran <i>et al</i> <sup>[38]</sup> , 2011	Lung/3DCRT,SBRT	PBC, Eclipse AAA, Pinnacle CCC, Masterplan PBC and CCC	PBC yielded higher TCP in comparison with other algorithms. For small tumor, TCP was overestimated by 4%-13% by PBC; for large tumor, there was an increase of up to 6%-22%
Liu <i>et al</i> <sup>[39]</sup> , 2013	Lung/SABR	EPL, MC	EPL overestimates dose by amounts that substantially decrease TCP in a large proportion. Compared with MC, prescribing based on EPL translated to a median TCP decrement of 4.3% (range, 1.2%-37%) and a > 5% decrement in 46% of tumors
Bufacchi <i>et al</i> <sup>[33]</sup> , 2013	Prostate, HN, Lung, Breast /3DCRT	PBC, AAA	NTCP calculated with AAA was lower than the NTCP calculated with PBC, except for the breast treatments
Chetty <i>et al</i> <sup>[30]</sup> , 2013	NSCLC/SABR	EPL-1D, EPL-3D, AAA, CCC, Acuros, MC	Average TCP decrements (5%-10%, ranging up to approximately 50%) were observed with model-based algorithms relative to the EPL-based methods

Eclipse AAA: Eclipse Anisotropic Analytical Algorithm; OTP CC: Oncentra MasterPlan Collapsed Cone algorithm; Pinnacle CC: Pinnacle Collapsed Cone algorithm; XiO Sup: XiO Multigrid Superposition algorithm; OTP PB: Oncentra MasterPlan Pencil Beam algorithm; XiO FFT: XiO Fast Fourier Transform Convolution algorithm; EPL: Equivalent path length; MC: Monte Carlo.

PTV minimum doses observed in the dose-volume histograms which were the direct products of the calculation algorithms. Though these differences did not have a significant effect on the PTV D95, they had a strong impact on the TCP. The results implied that more advanced algorithms are essential to assess the quality of the treatment clinically in the more realistic way. The detailed results of the percent mean tumor control probability (TCP) values for all algorithms as a function of PTV volume are cited and listed in Table 1.

In Bufacchi *et al*<sup>[33]</sup>'s study, the focus was shifted to the clinical implication of algorithms on NTCP models for four tumor sites: prostate, head and neck, breast and lung. The pencil beam convolution and anisotropic analytical algorithm were used for 80 treatment plans. The authors found that when the original PBC treatment plans were

recalculated using AAA with the same number of monitor units, the NTCP became lower, except for the breast treatments. Further the authors concluded that this difference in NTCP between PBC and AAA treatment plans could be clinically significant. In Nielsen *et al*<sup>[34]</sup>'s paper, the study was specifically focused on the influence of dose calculation algorithms on NTCP in NSCLC patients. Six dose algorithms from four different treatment planning systems were investigated: Eclipse AAA, Oncentra MasterPlan Collapsed Cone and Pencil Beam, Pinnacle Collapsed Cone, and XiO Multigrid Superposition and Fast Fourier Transform Convolution. NTCP values for heart and lungs were calculated using the relative seriality model<sup>[35]</sup> and the LKB model<sup>[36,37]</sup>, respectively. The authors found that the influence of density correction on the NTCP values depended on the dose calculation algo-

rithms and the NTCP model parameter set. Compared to mean lung dose (MLD) and V20, the calculated NTCP values for pneumonitis were more sensitive to the calculation algorithms. All these implied that for plan evaluation the algorithms play an extremely important role and the dosimetric parameters such as MLD and V20 might not be sensitive enough for the assessment. The differences of the quantities calculated with and without density correction using each algorithm are cited and listed in Table 2.

To summarize, we list the clinical impact of dose calculation algorithms in Table 3. Five references<sup>[30,33,34,38,39]</sup> with their results and conclusions are summarized.

## CONCLUSION

In this study we reviewed the commonly used dose calculation algorithms: correction-based type (1) methods and model-based type (2) methods. The calculation accuracy of different algorithms illustrated by several studies was summarized. Special focus was given to dose calculation comparison in the lung region. All the research studies demonstrated that for dose calculation in lung region, the advanced type (2) methods are necessary. Further, the accuracy of dose calculation algorithms was correlated to the quantities of TCP/NTCP, and the connection between the algorithms and clinical impact was established. The clinically related TCP/NTCP values are sensitive to the accuracy of dose algorithms. In conclusion, dose calculation algorithms play a vital role in radiation therapy.

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