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# Photon Dose Calculation Method Based on Monte Carlo Finite-Size Pencil Beam Model in Accurate Radiotherapy

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**Abstract.** This study mainly focused on the key technologies, the photon dose calculation based on the Monte Carlo Finite-Size Pencil Beam (MCFSPB) model in the Accurate Radiotherapy System (ARTS). In the MCFSPB model, the acquisition of pencil beam kernel is one of the most important technologies. In this study, by analyzing the demerits of the clinical pencil beam dose calculation methods, a new pencil beam kernel model was developed based on the Monte Carlo (MC) simulation and the technology of medical accelerator energy spectrum reconstruction. which greatly improved the accuracy of calculated result. According to the axial symmetry principle, only part of simulation results was used for the data of pencil beam kernel, which greatly reduced the data quantity of the pencil beam and reduced calculated time. Based on the above studies, the MCFSPB method was designed and implemented by the Visual C++ development tool. With several tests including the comparisons among the American Association of Physicists in Medicine (AAPM) No. 55 Report sample and the ion chamber measurement of lung-simulating inhomogeneous phantom in clinical treatment plan, the results showed that the maximum error of most calculated point was less than 0.5% in the homogeneous phantom and less than 3% in the heterogeneous phantom. This method met the clinical criteria, and would be expected to be used as a fast and accurate dose engine for clinic TPS.

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**Key words**: Accurate radiotherapy, dose calculation, pencil beam, Monte Carlo.

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# 1 Introduction

Dose calculation is one of the core functions in radiotherapy Treatment Planning System (TPS). International Commission on Radiation Units and Measurements (ICRU) NO.24 report [1] points out that the error of the primary focus' radical dose should be lower than 5%, otherwise the primary focus tumors will be out of control. There are two types of dose calculation methods [2–8]: analytic method and Monte Carlo (MC) method. Conventional analytic dose calculation method may result in large errors in the heterogeneous region; but in homogeneous region, analytic method may achieve good dose calculation results with high efficiency. Monte Carlo method may get the great accurate dose calculated result in both the homogeneous and heterogeneous region by simulating the transport of particles, but it is time-consuming in clinical usage [9–11].

In clinical TPS, pencil beam dose calculation algorithm was used widely. It is not easy to achieve pencil beam kernel database from measurement, with the reasons of small size field's accuracy and measured error. Monte Carlo method can get a more accurate field size and simulated result, but the photon spectrum is unknown. At the same time, the data quantity of conventional pencil beam kernel database is usually very big, which limited its usage.

Based on Monte Carlo code DOSXYZnrc in the EGSnrc system [12–14], a new pencil beam kernel model with the technology of energy spectrum reconstruction was developed. The photon dose calculation based on the Monte Carlo Finite-Size Pencil Beam (MCFSPB) meets the clinical needs, and will be expected to be used as a fast and accurate dose engine for Accurate Radiotherapy System (ARTS) [10,11]. ARTS is a comprehensive radiation treatment system supporting 3D Conformal Radiotherapy (3D-CRT), Intensity Modulated Radiotherapy (IMRT), Image Guided Radiotherapy (IGRT) and Dose Guided Radiotherapy (DGRT). With ARTS, clinical doctors and physicians can efficiently make, choose and verify the most suitable treatment plan. The flexibility and high efficiency features of ARTS provide specific treatment plans for different patients.

# 2 Methods

Finite-Size Pencil Beam (FSPB) method is one of most popular dose calculation methods in recent two decades. According to the idea of finite size pencil beam dose calculation [2], the dose of one point $r$ can be acquired by:

$$D(r)=\int_E \sum_s \phi_E(s)\Pi(E,r,s)dE, \tag{2.1}$$

where $\phi$ is the photon flux of the energy bin $E$ at the incident point $s$. $\Pi$ is the pencil beam kernel of the energy bin $E$ at the incident point $s$ and the calculated point $r$. The dose of the point $r$ is the sum of the incident points and the energy bins.

Considering the flux is very little different in the whole field when medical linac with a flattening filter, and the angle distribution of high-energy X-ray beam can be ignored in the range of whole field [15], the integration of energy spectrum can be included into the pencil beam kernel. Considering about energy spectrum reconstructed by measured percentage depth dose (PDD) [16], a new pencil beam method (Monte Carlo Finite-Size Pencil Beam, MCFSPB) was developed by FDS team, which can be simulated by Monte Carlo particle transport simulation codes, such as EGSnrc. In MCFSPB, Eq. (2.1) can be simplified as:

$$D(r)=\sum_{s}\phi(s)\Pi(r,s), \tag{2.2}$$

where $\phi$ is the photon flux of the reconstructed energy spectrum of linac at the incident point $s$. $\Pi$ is the pencil beam kernel of the reconstructed energy spectrum of linac at the incident point s and the calculated point $r$. The data of MCFSPB kernel is part information of only one standard vertical-incidence unit-field with the size of 0.5cm$\times$0.5cm, and some corrected technologies are used to calculation with any condition.

Consider the effect of particles' transport in the inhomogeneous organ of the patient and phantom, MCFSPB coupled with the Batho inhomogeneous correction method [6, 17], which had been used in clinic widely. So a more precise dose calculation result could be acquired by:

$$D(r)=\sum_{s}\phi(s)\Pi(r,s)CF(r), \tag{2.3}$$

where $CF(r)$ is the correct factor of the Batho method.

### 2.1 Photon energy spectrum reconstruction

The energy spectrum of high energy X-ray greatly affects the accuracy of radiation dose calculation. To obtain the photon spectra of medical linac in radiotherapy effectively, an analytical nonlinear programming model based on Monte Carlo PDD and measured PDD was investigated [16], and several regression algorithms including Levenberg-Marquardt, Quasi-Newton, Newton, Principal-Axis and Nminimize algorithms were used to realize this model, while the results were also compared with the conventional discrete method.

### 2.2 Pencil beam kernel acquisition

In finite size pencil beam dose calculation, the clinical beam is divided into some pencil beams and the patient body is divided into a 3D matrix of divergent calculation voxels. The acquisition of pencil beam kernel is key technology for MCFSPBmethod.

Monte Carlo code DOSXYZnrc in the EGSnrc system was used to simulate the special polyenergetic pencil beam energy deposition kernel. The phantom is a 30cm$\times$30cm$\times$30cm water (1.00g/cm$^3$) tank, with $300\times300\times300$ voxels. The origin of the coordinates (0,0,0) is placed at the geometrical central point of upper surface of the phantom. Along the direction of $X$ axis, phantom locates between [$-$15.0cm, 15.0cm], along the direction of $Y$ axis, phantom between [$-15.0$cm, 15.0cm], and along the direction of $Z$ axis, phantom between [0.0cm, 30.0cm]. The accelerating potential energy with the spectra reconstructed by above method, is simulated to irradiate the phantom with 0.5cm$\times$0.5cm rectangle open field, the radioactive source locates at $(0,0,-100)$, and the irradiation direction is parallel to the $Z$ axis and the central axis passes through the origin of coordinates $(0,0,0)$. Considering about axis symmetry of the phantom's dose distribution, part information of DOSXYZnrc's output file is acquired to make pencil beam energy deposition kernel.

# 3 Benchmark

Dose calculation is one of the core functions in TPS. In this paper, American Association of Physicists in Medicine (AAPM) No. 55 report [18] test cases and clinical examples were used to test MCFSPB dose calculation method.

## 3.1 AAPM benchmark

No. 55 Report of AAPM providing a set of complete dose data for the verification of the external photon beam algorithm, has been used to test dose calculation accuracy in TPS. There are 27 examples in the No. 55 Report, which include: particle energy, beam shape and size, source-skin distance change, wedge, block and tissue inhomogeneity as well as other factors that would affect the accuracy of dose calculation.

According to the situations (Table 1) shown in the report, the calculation results of interested points were compared with measured results from the report to verify the accuracy of the method.

The calculated points with the calculation error less than 3% are more than 98.8% in all calculated points, and the average errors were shown in Table 1. There are some situations with average error about 2%, such as large field size test case, "central block" test case of 18MV, "lung inhomogeneity" test case. The reasonable factors of large error will be discussed in the fourth part of this paper.

*Note:* Except special declaration clause, all other test cases are in the condition of vertical incidence to water phantom with the 100cm SSD for 18MV and 80cm SSD for 4MV.

## 3.2 Compared with measured result

In the clinic, there are many other factors would affect the accuracy of the method, such as the leakage of multi-leaves collimator (MLC for short), which is not included in No. 55 Report of AAPM.

To further verify the accuracy of MCFSPB dose calculation method in the clinical application, a lung-simulating inhomogeneous phantom (Fig. 1) was used. With the com-

Table 1: AAPM #55 report benchmark situations and results (4MV).

| Describe of the Example (VI:vertical-incidence) | Dose of Central Axis | | Dose of Inside Beam | | Dose of Outside Beam | | Width of Axis Field | |
|---|---|---|---|---|---|---|---|---|
| | Sampling point number | Average error (%) | Sampling point number | Average error (%) | Sampling point number | Average error (%) | Sampling point number | Average error (cm) |
| 5×5,VI,water | 8 | 0.225 | 8 | 0.263 | 8 | 0.550 | 8 | 0.015 |
| 10×10,VI,water | 8 | 0.225 | 8 | 0.500 | 8 | 0.913 | 8 | 0.009 |
| 25×25,VI,water | 8 | 0.350 | 8 | 1.150 | 8 | 2.088 | 8 | 0.031 |
| 5×25,VI,water | 8 | 0.475 | 8 | 0.588 | 8 | 1.013 | 8 | 0.039 |
| 25×5,VI,water | 8 | 0.725 | 8 | 0.850 | 8 | 0.75 | 8 | 0.046 |
| 10×10,VI,water, 70cm-SSD | 8 | 0.875 | 8 | 1.313 | 8 | 1.400 | 8 | 0.045 |
| 9×9,VI,water, 45-degree wedge | 5 | 0.140 | 16 | 0.556 | — | — | 8 | 0.055 |
| 16×16,VI,water, central-block | 8 | 0.838 | 8 | 0.563 | — | — | — | — |
| 10×10,VI,water, off-center plane | 8 | 0.300 | 8 | 0.438 | 8 | 1.100 | 8 | 0.041 |
| 16×16,VI,water, Irregular plane | 7 | 0.414 | 7 | 0.614 | — | — | — | — |
| 16×16,VI,lung-inhomogeneity | 5 | 1.940 | 10 | 0.93 | — | — | — | — |
| 16×16,VI,bone-inhomogeneity | 7 | 1.014 | 7 | 0.943 | — | — | — | — |
| 10×10,45-degree oblique-incidence,water | 6 | 0.800 | 12 | 1.183 | — | — | — | — |

parison of target (red point in Fig. 1) radiation dose value between calculation by MCF-SPB and measurement by ionization chamber, the results (Tables 2 and 3) showed that the target dose variation less than 3% in the three-dimensional conformal plan and less than 5% in the IMRT plan.

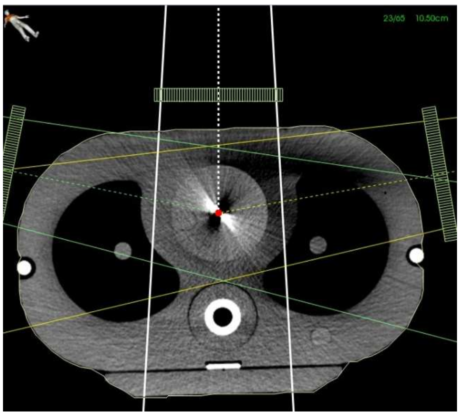

Figure 1: CT slice of target point in phantom.

Table 2: AAPM #55 report benchmark situations and results (18MV).

| Describe of the Example (VI:vertical-incidence) | Dose of Central Axis | | Dose of Inside Beam | | Dose of Outside Beam | | Width of Axis Field | |
|---|---|---|---|---|---|---|---|---|
| | Sampling point number | Average error (%) | Sampling point number | Average error (%) | Sampling point number | Average error (%) | Sampling point number | Average error (cm) |
| 5×5,VI,water | 8 | 0.263 | 8 | 0.413 | 8 | 0.675 | 8 | 0.024 |
| 10×10,VI,water | 8 | 0.225 | 8 | 0.350 | 8 | 0.688 | 8 | 0.021 |
| 25×25,VI,water | 8 | 0.313 | 8 | 1.025 | 8 | 3.000 | 8 | 0.026 |
| 5×25,VI,water | 8 | 1.3 | 8 | 0.775 | 8 | 1.288 | 8 | 0.041 |
| 25×5,VI,water | 8 | 0.263 | 8 | 0.575 | 8 | 0.763 | 8 | 0.080 |
| 10×10,VI,water, 85cm-SSD | 8 | 1.250 | 8 | 0.775 | 8 | 1.838 | 8 | 0.041 |
| 9×9,VI,water, 45-degree wedge | 8 | 0.325 | 16 | 0.531 | — | — | 8 | 0.054 |
| 16×16,VI,water, central-block | 8 | 2.875 | 8 | 1.063 | — | — | — | — |
| 10×10,VI,water, off-center plane | 8 | 1.188 | 8 | 1.163 | 8 | 0.825 | 8 | 0.038 |
| 16×16,VI,water, Irregular plane | 8 | 0.788 | 8 | 0.975 | — | — | — | — |
| 16×16,VI,lung-inhomogeneity | 7 | 1.386 | 14 | 1.286 | — | — | — | — |
| 6×6,VI,lung-inhomogeneity | 7 | 0.900 | 7 | 2.286 | — | — | — | — |
| 16×16,VI,bone-inhomogeneity | 7 | 0.343 | 7 | 0.900 | — | — | — | — |
| 10×10,45-degree oblique-incidence,water | 8 | 0.963 | 16 | 1.006 | — | — | — | — |

Table 3: Target dose of CRT test results.

| Case | Prescription | Field number | Beam angle() | | | MU | | | Measured | Error(%) |
|---|---|---|---|---|---|---|---|---|---|---|
| 1 | 200cGy | 3 | 0 | 80 | 281.4 | 81 | 107 | 102 | 198.14cGy | 0.94 |
| 2 | 200cGy | 5 | 0 | 45 | 100 | 51 | 59 | 61 | 205.16cGy | -2.52 |
| | | | 269 | 315 | — | 57 | 56 | — | | |

Table 4: Target dose of IMRT test results.

| Prescription | Field number | Beam angle() | Segment | MU | | | Measured | Error(%) |
|---|---|---|---|---|---|---|---|---|
| 201cGy | 3 | 0 | 1 | 182.3 | | | 209.1cGy | -3.87 |
| | | 83.7 | 3 | 61.22 | 45.37 | 25.57 | | |
| | | 283.1 | 5 | 56.98 | 48.36 | 34.18 | | |
| | | | | 18.82 | 18.17 | — | | |

# 4 Discussions

Flattening filter is invented to change beam off-axis intensity, in order to get a good off-axis ratio (OAR for short) at a depth. For some linac, the influence of flattening filter cannot be ignored. MCFSPB method did not consider the effect in the physics model at the beginning, so an off-axis correction with measured data was used to take account of flattening filter, which was not a very accurate correction. That is one reason to explain large error in some test cases. Intensity correction is considered to join the MCFSPB physics model, in order to improve the precision of calculated result.

Although MCFSPB method considered electronic contamination in the spectrum reconstruction, but the limitations of finite size pencil beam dose calculation method itself, which cannot accurately calculate the dose distribution in the central block test case with the influence of the linac head scattered radiation and electron contamination. This is one of reasons to explain the large error the central axis of the central block case in No. 55 Report of AAPM.

Batho method is one of the most popular inhomogeneous correction methods in modern TPS, with the characteristics of fast calculated speed, high precision at region of density changed smoothly and large error at region of density changed strongly. As a 1-dimension local correction method, it's fit for layered phantom, and hard for complex phantom or clinical patient. A 3-dimension un-local correction method is under development now, which will be more accurate to calculate complex phantom or clinical patient.

Beside the reasons mentioned above, there is another reason to explain the difference between calculated and measured results, which is the influence of MLC. MLC is used to change the shape of field, with widely application in accurate radiotherapy. Because of the difference between MLC's size and pencil beam size, and the leakage of MLC, there may be a large error between calculated and measured results. Intensity correction mentioned above will also consider about the leakage of MLC, in order to improve the precision of calculated result.

## 5 Conclusions

With the tests of the AAPM cases and three clinical cases, the MCFSPB method met the clinical criteria, and bad been used as a fast and accurate dose calculation engine for ARTS. Some advanced functions, such as point energy deposition kernel calculation method, are under development.

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