Monte carlo study on 6 MV photon beams of a CyberKnife® stereotactic radiosurgery system

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Abstract In this paper, the beam quality and percent depth dose curves for different field sizes of CyberKnife® system were investigated by Monte Carlo simulations using the PENELOPE code, which has been used to simulate 6 MV photon beam. In water phantom, the absolute doses were calculated for Φ10–60 mm collimators, and percent depth dose curves were evaluated for Φ30–60 mm collimators. The agreement of dose distributions of the calculation with measurement was within 3.0%. The mean energy of photon spectrum was 1.46 MeV, and the beam quality index was 0.632, which was slightly smaller than that of measurement.

Key words Monte Carlo, PENELOPE, CyberKnife, Beam quality, Percent depth dose

1 Introduction

The CyberKnife stereotactic radiosurgery system was developed jointly by Accuray Inc. and College of Medicine, Stanford University [1]. It is installed with a compact linac of 6 MeV electron beam on a robotic arm, a fix primary collimator system, and twelve final circular collimators. It has remarkable advantages in tracking the target positions of a patient, including guidance technique of advanced image, with great flexibility and accuracy [2–3], providing a highly conformal dose for the target while sparing nearby critical structures [4]. The single or multiple beams of nominal diameter of 5–60 mm are formed at 80 cm source-to-surface distance (SSD) from the upstream surface of a target, and are used to treat pathological changes with sub-mm accuracy in complex patient geometries, such as head, lung and neck [4]. Treating such a planning, however, needs an accurate dose calculation algorithm.

Monte Carlo (MC) technique can solve complex physical and mathematical problems because of its random sampling and high accuracy, whereas a simulation using conventional numerical methods with multiple independent variables demands formidable amounts of memory and computer time[5]. Radiation transport and dose distributions have been extensively studied by the MC technique in radiotherapy [6–9], e.g. PENELOPE [5], MCNP [10], EGS4 [11], and Geant4 [12].

In this work, beam quality and percent depth dose (PDD) were simulated for different field sizes at a CyberKnife® center. The calculated beam quality and PDD for 30–60 mm collimators at 80-cm SSD were compared with the results measured with a diode detector. The photon mean energy was discussed.

2 Methods and materials

The CyberKnife® system was installed at Hua Shan Hospital Affiliated to School of Medicine of Fu Dan University. All measurements were performed in a motorized water phantom (MP3, PTW, Freiburg, Germany) using the whole set of collimators ranging from 0 to 30 cm. We used PENELOPE-2006 [5], with which the transportation of electrons, positrons and photons in energy range of 100 eV–1 GeV in various kinds of materials can be simulated.

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Fig.1 shows schematically the treatment head of the CyberKnife® system. Geometry and materials of the simulation were chosen according to specifications of the linac. Monoenergetic incident electron beam of 6.0 MeV was used. The SSD was 80 cm from upstream target to the water phantom surface. The phase space plane was located on the top of second collimator, and was repeatedly used as inputs. The charge, energy, position, angle, and weight of particles were obtained from the phase files. For collimators of different sizes, the PDD curves were calculated by modeling each second collimator and the water phantom, and were measured by T60008 diode detector of 1-mm^2 cross section and 2.5-μm active layer (PTW, Freiburg, German), which was vertically positioned in isocentric setup.

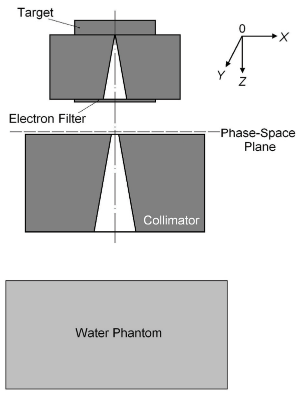

Fig.1 Schematics of the treatment head in the simulations. The scored phase-space plane is located on the top of second collimator.

In radiotherapy, parameters of the X-rays depend on energy of the E-beam, the beam quality was described as the transmission capability, which was expressed with tissue phantom ratio (*TPR*) in the 6 MV CyberKnife® system, and was defined as [13].

$$TPR = D_Q/D_{Q\mathrm{ref}} \tag{1}$$

where, D_Q and $D_{Q\mathrm{ref}}$ were the dose at an arbitrary point (Q) and a reference depth (Q_{ref}) in the water phantom along the beam central axis [14]. With an SDD of 80 cm, TPR_{20} and TPR_{10} represent the *TPR* at depths of 20 and 10 cm, respectively.

For the 6 MV linac, the beam quality index is,

$$TPR^{20}{}_{10} = TPR_{20}/TPR_{10} \tag{2}$$

where, TPR_{10} and TPR_{20} were obtained by simulating at d_1=10 and d_2=20 cm from the target to the detector in the water phantom, respectively.

A total of 2×10^6 electrons incident on the target through collimators of 10, 20, 30, 40, 50 and 60 mm in diameter was simulated under identical conditions. $TPR^{20}{}_{10}$ was measured with a waterproof Farmer ionization chamber of 0.6 cm^3 (TW30013, PTW, Freiburg) at 20 and 10 cm depths, respectively. It was placed in the water phantom at 80 cm SDD for the Φ60 mm collimator.

The PENMAIN code controls the simulation tracks so as to keep score of the relevant quantities. All events of photon interactions were simulated in a chronological sequence, in which the electron and positron transport were simulated by a mixed procedure, i.e. hard interactions were simulated in detail, and soft interactions were described by multiple-scattering approaches. To ensure reliable CPU time, the splitting idea and interaction force were adopted except for Russian roulette. The parameters are shown in Table 1.

Table 1 Simulation parameters of PENELOPE

Parameter	For all material
Eabs(γ) /eV	1.0×10^3
Eabs(e$^-$/e$^+$) /eV	1.0×10^4
C1	0.05
C2	0.05
Wcc /eV	1.0×10^4
Wcr /eV	1.0×10^3

3 Results and discussion

Fig.2 shows the absolute doses for collimators of 10, 20, 30, 40, 50 and 60 mm in diameter calculated using 6 MV photon beams. The dose decreases with the field size in water phantom. Therefore, the tumor size can be matched with different fields. This was the most advantageous to the clinic physicists.

Fig. 3 shows an energy spectrum at the phase-space plane. The photon energies are peaked at 0.52 MeV, with a mean energy of 1.46 MeV. This is slightly lower than the mean energy of 1.55 MeV by of Yamamoto *et al* [7], who used a thicker electron filter than ours. Our result is lower, too, than the mean energy (1.72 MeV) in Ref.[4], where the electron filter was made of lead, instead of aluminum in this study. These indicate that both the material and its thickness of the electron filter affect the photon energy spectrum significantly.

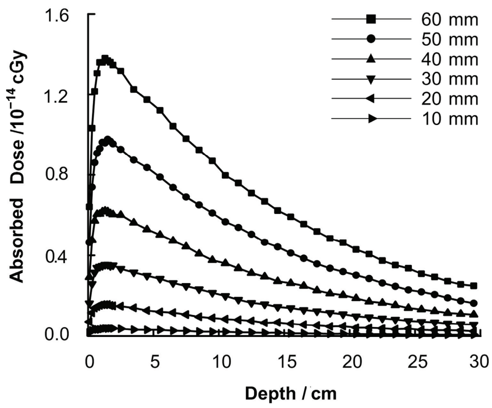

Fig.2 Absolute doses for collimators of 10, 20, 30, 40, 50 and 60 mm in diameter.

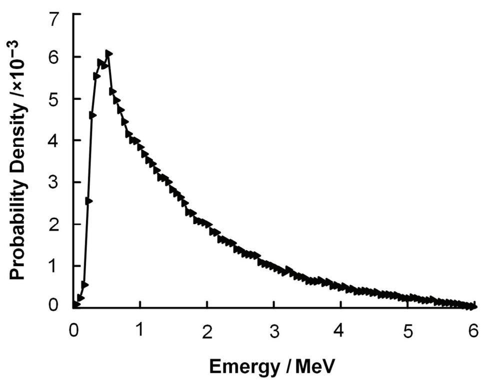

Fig.3 Energy spectrum of 6 MV photon beam from CyberKnife system.

In clinic, the oncologists may mainly consider target dose rather than the beam quality, but the beam quality is important for quality assurance of radiotherapy. The beam quality can be expressed in TPR^{20}_{10}, which is 0.632–0.644 in the protocol of American Association of Physicists in Medicine[15]. The TPR^{20}_{10} we calculated was 0.632, while the measured TPR^{20}_{10} was 0.640, both satisfying the QA requirement of the CyberKnife® system. The X-ray energy has influence on therapeutic effect, it is important to verify, frequently, the accelerator energy.

Fig.4 shows the measured and calculated PDD values for collimators of 30, 40, 50 and 60 mm in diameter. The PDD measurements were carried out at 1.5-cm depth on the beam central axis. It can be seen

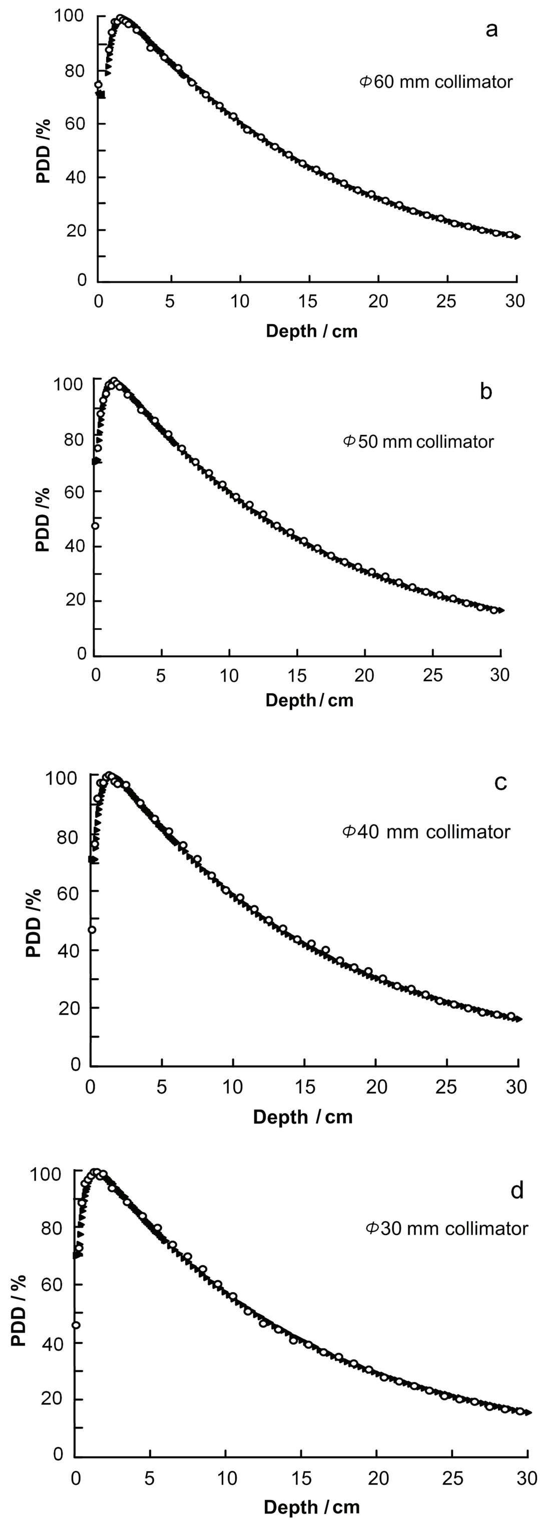

Fig.4. Measured (►) and calculated (○) PDD values for collimators of different diameters, with SSD = 80 cm.

that the measured dose distributions in the water phantom agree well (within 3.0%) with the calculated ones for various collimator sizes. The accuracy requirement on dose delivery in radiotherapy is about 5% [16,17]. This can be the result of an uncertainty in dose response. The calculated depth of maximum dose was d_{max}=1.4cm, while it was measured as d_{max}=1.5 cm. This is good enough, referring to the national standard criterion that the discrepancy of beam penetrability should not exceed 3 mm [18]. The main cause of this result was the low photon mean energy. As shown in Fig.2, the surface absorbed doses were less than 47 percent for all fields on the central axis in the buildup region. This also agreed with the 60% maximum dose of the national standard on the center of axis [19].

4 Conclusion

The simplified PDD curves and the absolute doses for various collimators were carried out by MC simulation on the beam central axis, and its agreement of the calculation with the measurement was within 3.0% in water phantom. The calculated $TPR^{20}{}_{10}$ was slightly lower than that of measurement. The thickness and material of the electron filter significantly influence on the photon mean energy. Our results showed the MC simulations may be effective and feasible for dose calculations, and beneficial to improve the accuracy and speed of radiotherapy QA for CyberKnife® system. The unresolved discrepancy, two-dimensional, and three-dimensional dose distributions need be further investigated in real case due to the limitations of the PENELOPE simulation and our measurements.

References

1 Adler J R, Chang S D, Murphy M J, *et al.* Stereotact Funct Neurosurg, 1997, **69:** 124–128.

2 Shiomi H, Inoue T, Nakamura S, *et al.* Radiat Med, 2000, **18(2):** 107–113.

3 Chang S D, Main W, Martin D P, *et al.* Neurosurg, 2003, **52:**140–146.

4 Deng J, Guerrero T, Ma C M, *et al.* Phys Med Biol, 2004, **49**:1689–1704.

5 Salvat F, Fernández-Varea J M, Sempau J. PENELOPE-2006: A Code System for Monte Carlo Simulation of Electron and Photon Transport, Barcelona, 2006, 4–7.

6 Fujio A. Med Phys, 2006, **33**(8): 2955–2963.

7 Yamamoto T, Teshima T, Miyajima S, *et al.* Radiat Medic, 2002, **20(6):** 305–310.

8 Lymperopoulou G, Petrokokkinos L, Papagiannis P, *et al.* Nucl Instrum Meth Phys Res A, 2007, **580**: 548–551.

9 Castellano G, Brusa D, Carrara M, *et al.* Nucl Instrum Meth Phys Res A, 2007, **580**: 502–505.

10 MCNP - A General Monte Carlo N-particle transport code, http://mcnp-green.lanl.gov/

11 Distribution area for egs4unix_3.0, http://www.irs.inms. nrc.ca /EGS4/get_egs4.html

12 Geant4 user's guide, http://geant4.web.cern.ch/geant4/

13 Andreo P, Burns D T, Hohlfeld K, *et al.* Technical Report Series 398: Absorbed dose determination in external beam radiotherapy: an international code of practice for dosimetry based on standards of absorbed dose to water, 2000, 190–388.

14 Podgorsak E B. Radiat oncolo phys, IAEA, VIENNA, Austria, 2005,190.

15 Almond P R, Biggs P J, Coursey B M, *et al.* The AAPM's TG-51 Protocol for Clinical Reference Dosimetry of High-Energy Beams, Med Phys, 1999, **26(9):**1847–1870.

16 ICRU Report-24: In Determination of Absorbed Dose in a Patient Irradiated by Beams of X or Gamma Rays in Radiotherapy Procedures, ICRU, Washington DC, 1976.

17 ICRU Report-42: In Use of Computers in External Beam Radiotherapy Procedures with High Energy Photons and Electrons, ICRU, Washington DC, 1987.

18 GB 15213-94, Medical electron accelerators-Functional performance characteristics and test methods, 1994. (in Chinese)

19 Kong X, Zhao H, Li Q, *et al.* High Ener Phys Nucl Phys 2005, **29(9):** 903–906. (in Chinese)