

A CT-based Monte Carlo Dose Calculations for Proton Therapy Using a New Interface Program

A. Esmaili Torshabi, A. Terakawa, K. Ishii, H. Yamazaki, S. Matsuyama, Y. Kikuchi, M. Nakhostin, H. Sabet, A. Ishizaki, W. Yamashita, T. Togashi, J. Arikawa, H. Akiyama, K. Koyata

Abstract—The purpose of this study is to introduce a new interface program to calculate a dose distribution with Monte Carlo method in complex heterogeneous systems such as organs or tissues in proton therapy. This interface program was developed under MATLAB software and includes a friendly graphical user interface with several tools such as image properties adjustment or results display. Quadtree decomposition technique was used as an image segmentation algorithm to create optimum geometries from Computed Tomography (CT) images for dose calculations of proton beam. The result of the mentioned technique is a number of non-overlapped squares with different sizes in every image. By this way the resolution of image segmentation is high enough in and near heterogeneous areas to preserve the precision of dose calculations and is low enough in homogeneous areas to reduce the number of cells directly. Furthermore a cell reduction algorithm can be used to combine neighboring cells with the same material. The validation of this method has been done in two ways; first, in comparison with experimental data obtained with 80 MeV proton beam in Cyclotron and Radioisotope Center (CYRIC) in Tohoku University and second, in comparison with data based on polybinary tissue calibration method, performed in CYRIC. These results are presented in this paper. This program can read the output file of Monte Carlo code while region of interest is selected manually, and give a plot of dose distribution of proton beam superimposed onto the CT images.

Keywords—Monte Carlo, CT images, Quadtree decomposition, Interface program, Proton beam

I. INTRODUCTION

PROTON therapy is one of the most effective methods of cancer treatment in radiotherapy because of the sharp penumbra and Bragg peak effect (Wilson 1946). In this method, the tumor volume is irradiated through a highly focused proton beam with a remarkable accuracy which reduces damages to the normal tissues. An effective treatment is achieved with high precision of dose calculation.

A. E. Torshabi, A. Terakawa, K. Ishii, S. Matsuyama, Y. Kikuchi, M. Nakhostin, H. Sabet, A. Ishizaki, W. Yamashita, T. Togashi, J. Arikawa, H. Akiyama and K. Koyata are with Department of Quantum Science and Energy Engineering, Graduate School of Engineering, Tohoku University, Sendai, Japan.

H. Yamazaki is with Cyclotron and Radioisotope Center, Tohoku University (corresponding author: phone/fax: +81227957933, e-mail: ahmad@stein.qse.tohoku.ac.jp).

In recent years, Monte Carlo simulation method has been known as the most accurate method of radiation transport and dose distribution calculations in complex heterogeneous systems such as human body organs. Procedures for simulating patient dose distributions are moving toward medical image processing and using image information to construct more realistic models of human body [1]-[2]-[3]. Geometry models derived from tomography images provide accurate representations of organs by discrete elements of volume called voxels. The voxel size is one of the key factors that affects on the Monte Carlo calculation accuracy and run time [4]. High accuracy is achieved with small voxel size but decreasing voxel size leads to highly time intensive which limits practical use of voxel-based simulation methods.

The aim of this study is to calculate the Monte Carlo (MCNPX-based) dose distribution of proton beam in complex heterogeneous systems such as organs or tissues via a new interface program independent of voxel based algorithms in image segmentation to improve dose calculations through geometry optimization.

Quadtree decomposition technique is used to segment images. This technique divides an image into four equal-sized square blocks, and then checks each block to see if it meets some criterion of homogeneity. If a block meets the criterion, it is not divided any further. If it does not meet the criterion, it is subdivided again into four blocks, and the test criterion is applied to those sub-blocks. The division can be continued with blocks as small as voxel size if it was needed. Finally the geometry of organ or tissue is represented by individual cells with different sizes. By this way the resolution of image segmentation is high enough in heterogeneous areas such as the interfaces between two medium to preserve the precision of dose calculations and is low enough in homogeneous areas to reduce the number of cells directly.

Although quadtree decomposition method reduces the number of cells directly, interface program also consists of a Cell Reduction Algorithm (CRA) in order to reduce the number of cells by joining neighbor cells having the same material. If whole image data is not needed for geometry construction, user can crop any arbitrary square region, interactively.

The validation of this interface program has been done in two ways. In first method a block-shaped PMMA phantom was used as target in front of 80 MeV proton beam in CYRIC

of the Tohoku University and the comparison between simulation and experimental results are presented in this work. In the second method a multilayer phantom with different materials near body tissues was used as target in simulation procedure and simulation data were compared with data on the base of the polybinary tissue calibration method [5], performed in CYRIC.

II. MATERIAL AND METHOD

A. Software Development

This interface program was developed under MATLAB 7.0 software package. A friendly Graphical User Interface (GUI) was created with several options. It allows user to enter inputs such as CT images, materials, source and target specifications, etc. Medical images with DICOM (Digital Imaging and Communications in Medicine) format can be read as an input file for interface program, although other common image formats are acceptable as well. By image review and geometry display options, user is able to review tomography images and evaluate final geometry produced by interface program in MATLAB plot. User can also determine proton beam specifications such as energy and beam type and target specification such as localization and direction. This program includes three different type of proton beam which are pencil beam, broad beam and Spread Out Bragg Peak (SOBP) with different uniform dose profiles in depth.

Pencil beam: The characteristics of pencil beam are energy and FWHM (Full Width at Half Maximum). It should be considered the atomic compositions of the air which exist in beam path causes a few scattering on proton beam with a narrow Gaussian profile. In simulation, user is able to change Energy and FWHM of pencil beam to achieve desired dose distribution inside target. In interface program, a proper window was made for user to enter energy and FWHM of pencil beam.

Broad beam: Broad beam is created by inserting a Lead foil with 0.1 mm thickness in front of proton beam. After irradiating protons to scatterer, the narrow pencil beam will be expanded into a larger beam with an approximately Gaussian transverse profile. Energy and FWHM of this type of proton beam can be entered via a related window of Interface program and their values depend on the user plan about dose distribution inside target.

SOBP: Spread out Bragg peak is created using ridge filters in CYRIC. Different type of ridge filters makes SOBP with different flat area of dose profile in depth. All types of ridge filters were defined in simulation part and user is able to select desired ridge filter to obtain appropriate SOBP inside target.

Quadtree decomposition technique is used as image segmentation algorithm and the resolution of segmentation or the homogeneity test factor is variable which is determined by user for geometry construction. A segmented image has been shown in Fig. 1 using Quadtree decomposition technique in image segmentation with a specific homogeneity test factor.

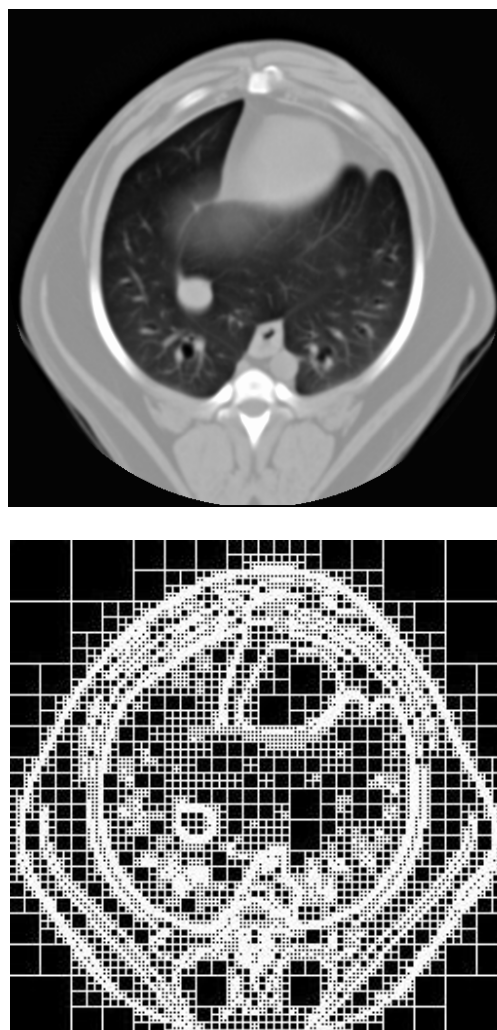


Fig. 1 A CT image (Top) and Segmented image (Bottom)

User can also apply CRA to reduce the number of individual cells in MCNPX input file in order to decrease MCNPX execution time. By applying CRA same geometry is created with fewer cells in comparison with normal geometry. Moreover, Image cropping option allows user to crop any arbitrary square region interactively, while whole image pixels are not needed for geometry construction. By applying this option, pixels outside of cropped region are not used in MCNPX input file generation. Material densities and their atomic compositions have been defined from International Commission on Radiation Units and measurements (ICRU) report No. 44 [6]. Material data can be adjusted by operator.

The interface program can read output file of MCNPX while Region of Interest (ROI) is selected manually by user, and give a plot of dose distribution of proton beam superimposed onto the tomography images. Fig. 2 shows a block diagram of this interface program.

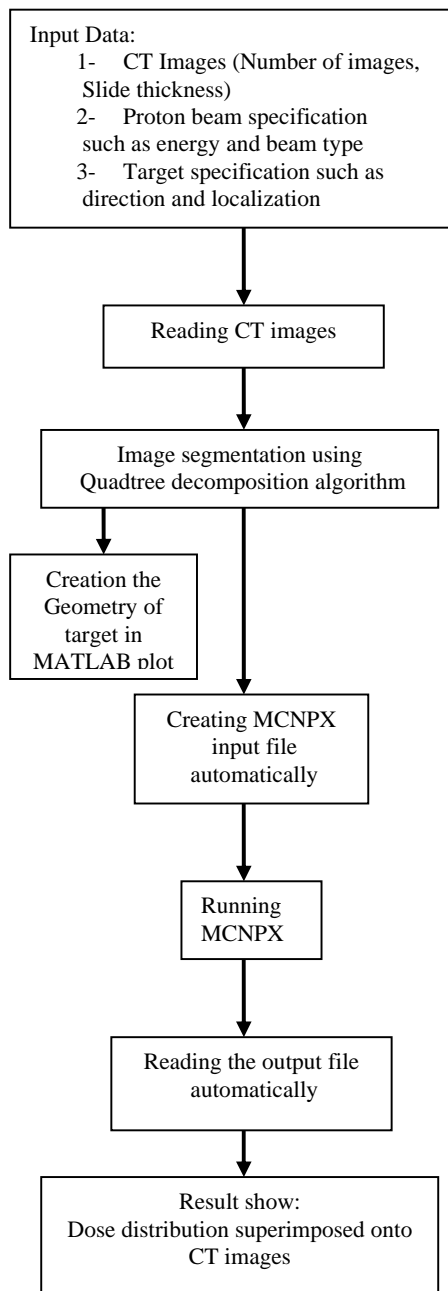


Fig. 2 Block diagram of interface program

B. MCNPX Monte Carlo Code

The Monte Carlo method is a well-established technique that has found widespread uses in situations where physical measurements are either difficult or impossible. In the context of radiation transport, the technique involves the use of random numbers and probability distributions to determine the energy, position, direction and path-length of individual particles, as well as the type of physical interactions that particles experience when traversing a medium, and finally the energy deposited in defined volumes for the purpose of equivalent dose assessment. In this study calculations were carried out using MCNPX 26C Monte Carlo code developed

at the Los Alamos National Laboratory.

C. Experimental Setup

Experiments were performed in the CYRIC of Tohoku University. Two phantoms selected for this study are: A Polymethyl methacrylate (PMMA) block-shape phantom with 100mm×100mm×30mm size and a multilayer cylindrical phantom with 40 mm diameter and 40 mm length kept inside a holder with 75mm×75mm cross section size, as shown in Fig. 3. This phantom includes 8 layers with 5mm thickness in every layer and made by PMMA, flexible bone, muscle and water equivalent materials. The arrangement of layers has been shown in Fig. 3.

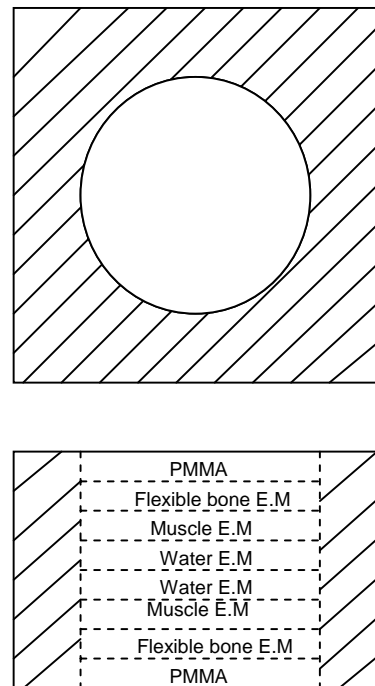


Fig. 3 A schematic layout of top view (top) and side view (bottom) of multilayer phantom (dash lines) with holder (solid lines). In this figure E.M. represents Equivalent Material

These two phantoms were submitted to Kitasato University for taking CT images to use for MCNPX simulation via interface program.

PMMA phantom was selected as irradiation phantom in experimental step. The irradiation source is 80 MeV proton beam produced by AVF (Azimuthally Varying Field) cyclotron. The smaller side of the phantom is adjusted vertically, in front of horizontal proton beam which has 365 cm distance from the exit window of proton beam. Experiments were carried out using two types of beam line with Spot and Broad shape. The dose distribution measurements were carried out using the Imaging Plate¹. Stopping power dependence [7] of Imaging Plate data was corrected with correcting factor obtained by Pin Point

¹ Imaging Plate (IP) made by Fuji Photo film CO. Ltd

Ionization Chamber² (Type 31014). These measurements comprise spot and broad beam profiles on X and Y axis and also depth dose distribution of proton beam inside phantom.

D. Simulation

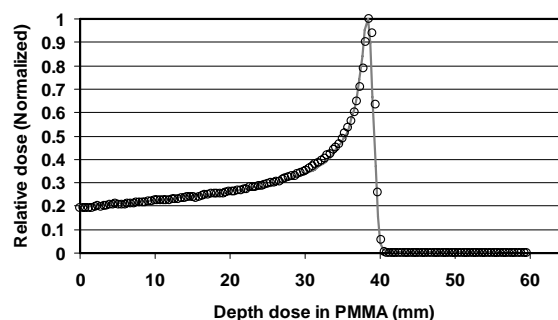
Monte Carlo simulations were accomplished using MCNPX. This code is increasingly popular in many medical applications such as radiotherapy treatment planning and dose assessment. Two recent efforts have been done in the Institute de Radioprotection et de Sûreté Nucléaire (IRSN) [1] and by Kenneth A. Van Riper [2] using MCNP.

Energy Deposition Mesh Tally (TMESH) and Energy Deposition Tally (F6) were used to perform calculations. Two modes of simulation have been considered using PMMA phantom. The first mode (simple mode) contains a phantom made by six surfaces in MCNPX input file with the same dimensions as PMMA phantom used in experimental method. In the second mode (CT-based mode) the same geometry of the PMMA phantom is made by interface program based on CT images took from phantom. Six CT images with 256×256 matrix size and 5 mm slice thickness were used to create the geometry. Atomic compositions and density of PMMA phantom have been entered via related material window of GUI. Other parameters are same for both modes in the simulation process, such as source specifications, phantom localization and direction. Due to negligible effect of the Imaging Plate thickness on simulation results, it has not been defined in simulation process.

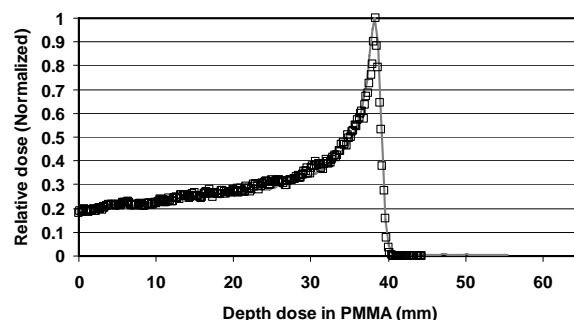
As a second step in simulation part, multilayer phantom CT data was used to construct its geometry. As shown in Fig. 3, the water equivalent plates were located in central part of cylinder and the purpose is irradiating Spread Out Bragg Peak (SOBP) proton beam with 10mm flat area to cover just water equivalent plates. For this aim, a 74 MeV proton beam was directed parallel to the cylindrical axis and was incident perpendicularly at the center of the circular cross section of this phantom. Simulation data obtained in this step was compared with data obtained on the base of the polybinary tissue calibration method [5], in CYRIC.

III. RESULT AND DISCUSSION

Fig. 4 represents simulated and measured results of depth dose distribution of proton beam inside PMMA phantom in the direction of the central axis of proton beam. As mentioned in simulation part, two modes of MCNPX input file have been defined for PMMA phantom: simple mode and CT-based mode and Fig. 4(a) shows simulated results between these two modes. Fig. 4(b) shows CT-based simulation and experimental results.



(a)



(b)

Fig. 4 Depth dose distribution of proton beam inside PMMA phantom, (a) MCNPX simulation Data in simple mode (circles) and CT-based mode (solid line), (b) Experimental (squares) and CT-based simulation Data (solid line)

The difference between simulation data in CT-based mode and simple mode is less than 5% and the difference between experimental data and CT-based simulation data is less than 8%. It can be noted these errors are associated to experimental effects such as energy fluctuation of proton beam which is $\pm 0.5\%$ MeV, Secondary particles, stopping power dependence of Imaging Plate and also Monte Carlo simulations. Nevertheless, there is a good agreement between simulation and experimental results.

Table I shows the energy deposition of 80 MeV proton beam (using scatterer) inside whole PMMA phantom as target in simple and CT-based simulation modes. Tally F6 is used to perform these calculations.

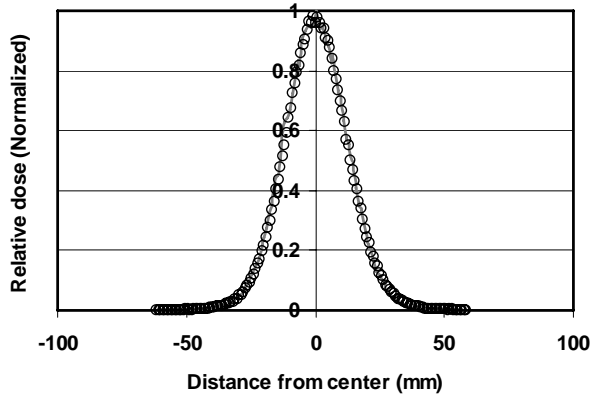
TABLE I
 ENERGY DEPOSITION IN PHANTOM (MEV)

Geometry type	Target Mass (g)	Energy Deposition (MeV)
Simple mode	357.0	17.230
CT-based mode	355.71	17.3102
CT-based mode (after applying CRA)	355.71	17.3105

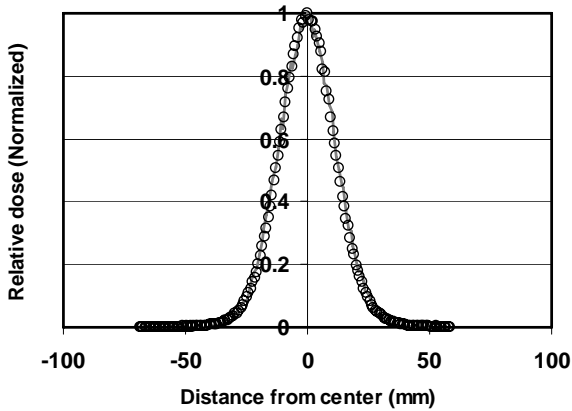
In two CT-based modes Energy Deposition in the phantom is calculated with summation of Energy Deposition of total cells which create phantom.

² PinPoint Ionization chamber Type 31014 made by PTW. Freiburg. An ultra small-sized therapy chamber with sensitive volume of only 0.015 cm³ and 4 mm in diameter

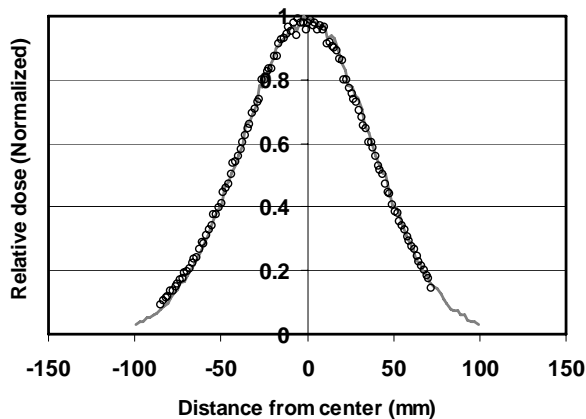
Transverse beam profiles of spot and broad beam were measured by Imaging Plate in 6 mm depth of PMMA phantom and calculated by MCNPX in the same depth. These results are shown in Fig. 5 on lateral X and Y directions.



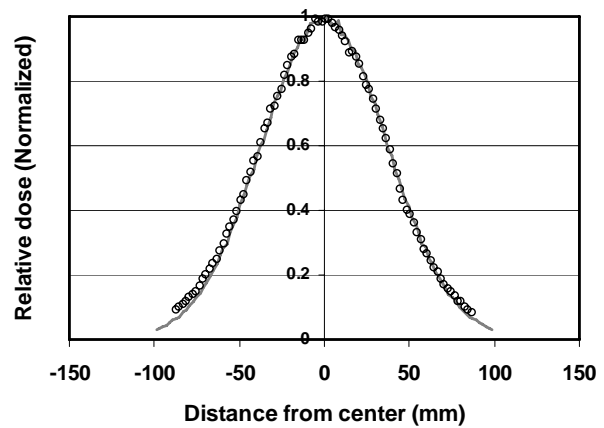
(a)



(b)



(c)



(d)

Fig. 5 Simulated (solid line) and experimental (circles) transverse dose profiles of spot and broad beam. (a) Spot beam profile on X-axis, (b) Spot beam profile on Y-axis, (c) Broad beam profile on X-axis, (d) Broad beam profile on Y-axis

The maximum differences between simulation and experimental values in spot beam profiles on X and Y-axis are less than 9% and 11% respectively. For broad beam profiles this difference is less than 3% for both, X and Y direction.

Fig. 6 represents the simulation results of depth dose distribution using CRA. In this case, by activating CRA, the interface program has generated two separate MCNPX input files of PMMA phantom geometry with different number of cells.

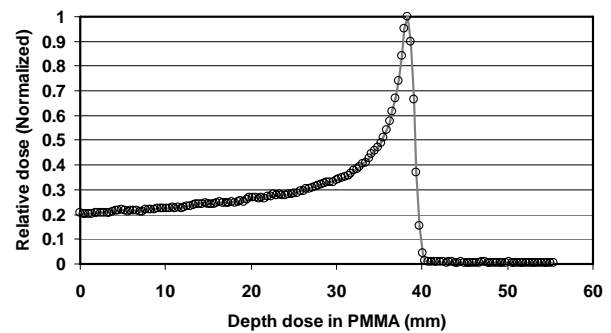


Fig. 6 Depth dose distribution of proton beam before (circles) and after (solid line) using CRA employment in Monte Carlo calculation

As seen, no significant difference between the two depth dose profiles is observed and hence CRA does not reduce dose calculations accuracy. It should be mentioned for this example run time improvement is 28% using CRA.

Fig. 7 represents the energy deposition of proton beam superimposed onto a CT image of PMMA phantom in a region of interest which is manually selected by user.

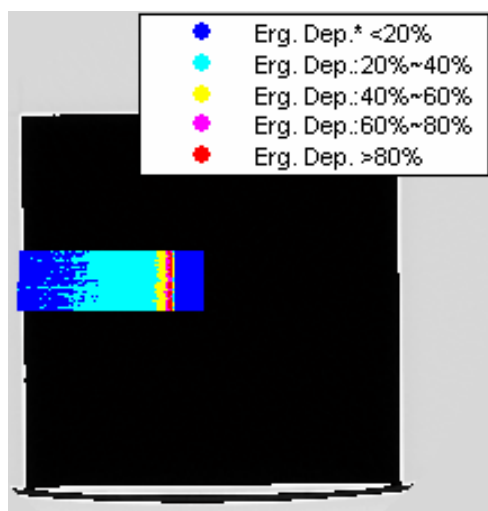


Fig. 7 Energy Deposition of proton beam superimposed onto phantom CT image.

* Erg. Dep. in above figure is abbreviation of Energy Deposition.

The penetration of proton beam inside phantom in Fig. 7 is equal to depth dose of proton beam shown in Fig. 4.

As explained in simulation part, multilayer CT data is used to simulate depth dose using SOBP proton beam with proper flat area to cover water equivalent materials. In this case, SOBP proton beam was formed by definition of ridge filter in MCNPX input file to give 10mm flat area. Fig. 8 shows the depth dose results inside this phantom. The difference between these data is 0.13mm in distal part of depth dose distribution.

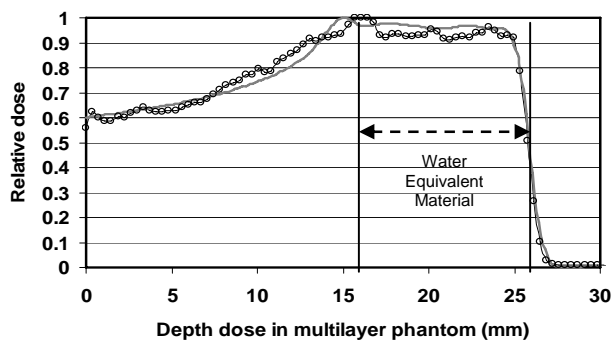


Fig. 8 Depth dose distribution of 74 SOBP (10mm) proton beam obtained by two methods: simulated by MCNPX (circles) and data based on polybinary tissue calibration method (solid line)

IV. CONCLUSION

A new interface program was introduced to make complex heterogeneous geometries preserving high accuracy in dose calculations and reducing execute simulation time. A friendly GUI has designed which allows user to process CT images and create MCNPX input file for proton therapy treatment planning. The capability of this program was demonstrated by an experimental and simulation arrangement using a block-

shaped PMMA phantom and a multilayer phantom for proton beam. The implementation of this interface program for more complex phantoms with different materials and also using human CT images are under process as a next plan with concentrating on dose distribution calculation of proton beam near boundaries between two mediums.

REFERENCES

- [1] S. Chiavassa, M. Bardies, F. Guiraud-Vitoux, D. Bruel, J. R. Jourdan, D. Franck, and I. Aubineau-Laniece, "OEDIPE: a personalized dosimetric tool associating voxel-based models with MCNPX," *Cancer Biother.*, vol. 20, no. 3, pp. 325–332, 2005.
- [2] K. A. Van Riper, "A CT AND MRI SCAN TO MCNP INPUT CONVERSION PROGRAM," *Radiat Prot. Dosim.*, vol. 115, no. 1-4, pp. 513–516, Dec. 2005.
- [3] C. Oliveira, H. Yoriyaz, F. C. Oliveira, and L. M. Ferreira, "Monte Carlo simulation for dose distribution calculations in a CT-based phantom at the Portuguese gamma irradiation facility," *Nucl. Inst. and Meth. in Phys. Res. B*, vol. 213, pp. 662–665, Jan. 2004.
- [4] W. Ai-dong, W. Yi-can, T. Sheng-xiang, and Z. Jiang-hui, "Effect of CT Image-based Voxel Size On Monte Carlo Dose Calculation," in *Proc. 27th Annu. Conf. Engineering in Medicine and Biology*, Shanghai, 2005, pp. 6449–6451.
- [5] N. Kanematsu, N. Matsufuji, R. Kohno, S. Minohara, and T. Kanai, "A CT calibration method based on the Polybinary tissue model for radiotherapy treatment planning," *Phys. Med. Biol.*, vol. 48, pp. 1053–1064, 2003.
- [6] International Commission on Radiation Units and Measurements, "Tissue Substitutes in Radiation Dosimetry and Measurement," *ICRU rep. no. 44* Maryland, Bethesda, 1989.
- [7] R. Kohno, A. Nohtomi, Y. Takada, T. Terunuma, T. Sakae, and K. Matsumoto, "A compensation method of an imaging plate response to clinical proton beams," *Nucl. Inst. and Meth. in Phys. Res. A*, vol. 481, pp. 669–674, 2004.