A Customized Cancer Radiation Treatment Planning Simulation (ccRTPs) System via Web and Network

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The telemedicine using independent client-server system via networks can provide high quality normalized services to many hospitals, specifically to local/rural area hospitals. This will eventually lead to a decreased medical cost because the centralized institute can handle big computer hardware systems and complicated software systems efficiently and economically. Customized cancer radiation treatment planning for each patient is very useful for both a patient and a doctor because it makes possible for the most effective treatment with the least possible dose to patient. Radiation planners know that too small a dose to the tumor can result in recurrence of the cancer, while too large a dose to healthy tissue can cause complications or even death. The best solution is to build an accurate planning simulation system to provide better treatment strategies based on each patient's computerized tomography (CT) image. We are developing a web-based and a network-based customized cancer radiation therapy simulation system consisting of four important computer codes; a CT managing code for preparing the patients target data from their CT image files, a parallel Monte Carlo high-energy beam code (PMCEPT code) for calculating doses against the target generated from the patient CT image, a parallel linear programming code for optimizing the treatment plan, and scientific data visualization code for efficient pre/post evaluation of the results. The whole softwares will run on a high performance Beowulf PC cluster of about 100−200 CPUs. Efficient management of the hardware and software systems is not an easy task for a hospital. Therefore, we integrated our system into the client-server system via network or web and provide high quality normalized services to many hospitals. Seamless communication with doctors is maintained via messenger function of the server-client system.

Key Words: Telemedicine, Client-server system, Customized radiation therapy planning simulation system, PMCEPT code, Beowulf PC cluster

INTRODUCTION

Customized cancer radiation treatment planning for each patient is very useful for both a patient and a doctor because it makes possible for the most effective treatment with the least possible dose to patient.[1] Radiation planners know that too small a dose to the tumor can result in recurrence of the cancer, while too large a dose to healthy tissue can cause complications or even death. However, accurate dosimetry modeling to match the detailed knowledge of the human body provided by CT data is not an easy problem. Moreover, technological revolution in radiation treatment planning is driving the need for more accurate dose calculations. The prevalence of three dimensional planning systems such as computer controlled multileaf collimators, image guided radiation therapy (IGRT), and 4-dimensional treatment (4D RT) including patient movement, require the ability to accurately correlate clinical outcome with dose and dose-volume information.[2] The value of such comparisons is questionable if the dose information is only approximate. In order to improve our treatment planning simulation accuracies, we need to pursue Monte Carlo modeling techniques (i.e.,

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PMCEPT code[3]) and parallel linear programming for optimizing the treatment plan.[4] The difficult issue is computing time when we use Monte Carlo and large dimensional linear optimization models. To overcome this difficulty, we must use a high performance super computer such as Beowulf PC cluster. Another difficulty is to use the given software efficiently. These two tasks may require a hospital to invest many resources because they require special expertise.

At this point, we suggest a client-server system via networks, a type of telemedicine. The European Commission's health care telematics programme defines telemedicine as a rapid access to shared and remote medical expertise by means of telecommunications and information technologies, no matter where the patient or relevant information is located. The center has high performance super computers and many resources of expertise to develop Monte Carlo and optimization codes and can provide customized cancer radiation therapy planning solutions to many hospitals, specifically to the local and rural area hospitals. Commercial secure audio-video teleconference system is used for the real time secure communication between the center and the member hospital. The member hospital will send a patient CT scan data to the center via networks. The patient CT data will be analyzed and converted to the target data including physical and chemical information. Accurate Monte Carlo and optimization codes will calculate the best planning solution within a reasonable time limit. The visualization software will produce graphics and pictures easy to be read by concerned members in the hospital. The details of concerning softwares will be described in the next section.

NEW CUSTOMIZED CANCER RADIATION THERAPY PLANNING SOLUTION SYSTEM

Commercial secure teleconference systems are now easily available everywhere. One server can have a few tens of client nodes that are mere personal computers with client software. Communication between server and client is realized in real time and securely. Specifically, Korean network system is one of the best in the world in reliability and speed.

Gigabits or sub gigabits networks are available to any local and rural area in Korea. In other words, Korea has the world best base infrastructure for telemedicine. We can take advantage of this technology to provide any rural area hospital with high quality customized cancer radiation therapy planning simulation solutions (CCRTPSS). The schematic model of the server-client planning system is shown in Fig. 1. We are developing a new CCRTPSS via networks. The CCRTPSS includes four important softwares that are described in detail in the following.

1. CT manager

For the customized planning, the first thing is to analyze the patient CT data file (usually, dicom format file) to get voxel

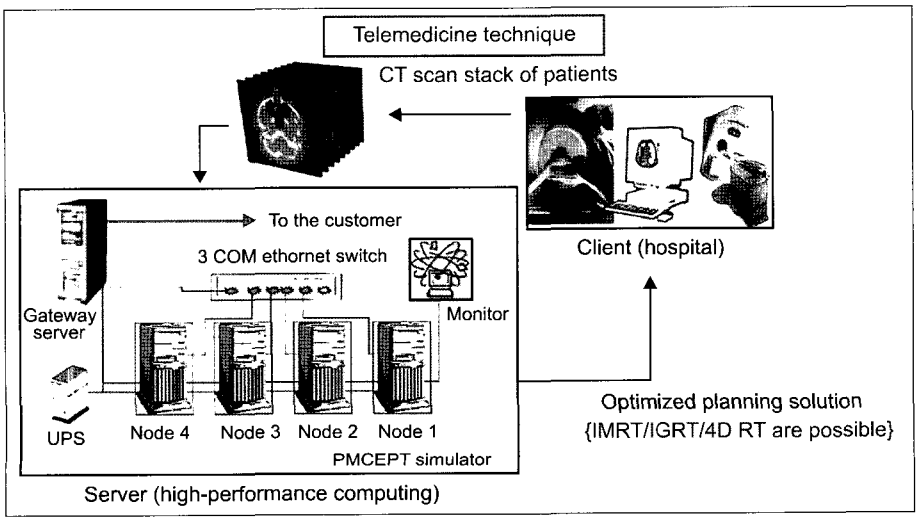

Fig. 1. The schematic model of the server-client planning simulation system. The hospital (client) sends a patient CT data to Server via networks. The high performance computers in the center calculate various complicate problems regarding IMRT/IGRT/4D RT simulations on time and provide the client with the best optimized solution in an easy-to-read graphical form.

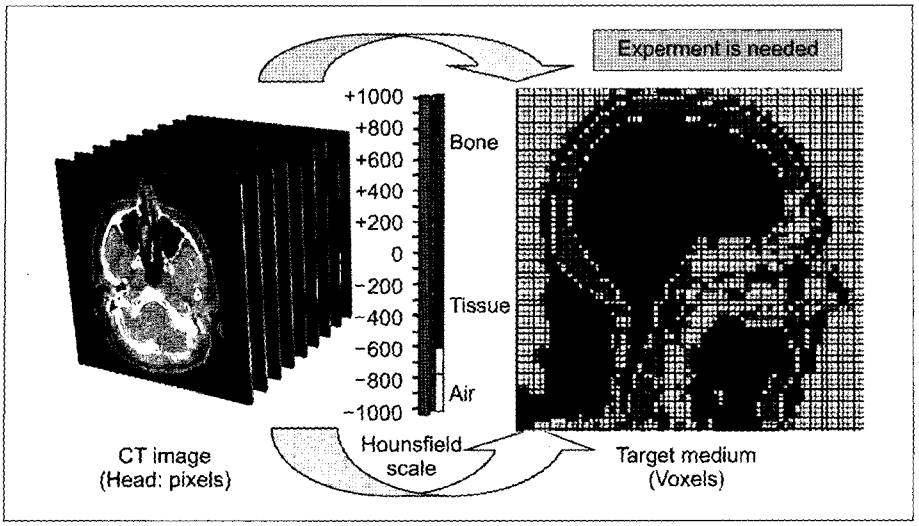

Fig. 2. The schematic picture to convert CT numbers from dicom files into mass density and elemental weight of tissues required as input for dose calculations with PMCEPT code. The pixel data of dicom file is converted into voxel data. Experimental adjustment is needed because most CT scanners have unknown distribution of polychromatic x-rays.

structure target data of mass density and elemental weight of atomic compositions in each voxel. Measuring projected transmission profiles of x-rays penetrating a patient from different directions producing the imaging in CT scanner. Thus, the CT number, called Hounsfield unit, is the attenuation coefficient of x-rays passing through the patient tissues. Much work to establish a relationship between CT number and tissue parameters has been performed. Among them, we used the recent Schneider *et al*'s[5] work in this system. They calculated the CT numbers for 71 human tissues in Ref. 5. The summary of equations for converting CT number into tissue parameters is provided in the Appendix 1. Fig. 2 shows the schematic picture to convert CT numbers into mass density and elemental weights of tissues required as input for dose calculations with parallel Monte Carlo electron and photon transport (PMCEPT) code.

2. Monte Carlo transport code

Most researchers who are pursuing more accurate dosimetry modeling are pursuing Monte Carlo modeling techniques in order to improve their treatment planning capabilities. It has been proved to be more accurate method than any semi-empirical and pencil beam methods for radiation treatment dose calculations. Specifically, a Monte Carlo method only account in a natural way for electronic nonequilibrium at medium interfaces. However, the difficult problem in Monte Carlo algorithms is that they are too slow to be acceptable for clinical purposes. This difficulty could be solved by either using faster computers or introducing new calculation techniques or both.

The PMCEPT code[3] is based on general-purpose Monte Carlo code for coupled electron and photon transport. It employs a message passing interface (MPI[6]) for the parallel computing. A brief overview of the physical and computational models is given in the Appendix 2. The computing time benchmark between PMCEPT and MCNP5[7] for the depth dose calculations in a water phantom irradiated with 20 MeV electron beam showed that the PMCEPT was approximately twenty times faster than the MCNP5 on the IBM laptop (ThinkPad X40 with 1.2 GHz cpu and 512 MB RAM memory) operated by the RedHat Linux 9.0. To validate the PMCEPT code, we have performed extensive benchmark tests by comparing PMCEPT results with absolute calorimetric dose measurement[8] for low incident kinetic energies of 0.5-MeV (left-hand panel in Fig. 3) and 1.0-MeV (right-hand panel in Fig. 3) electron beams incident on homogeneous and multi-layered media. In addition, comparisons between the PMCEPT and EGS4[9] and DPM[10] codes for high energy electron beam ranging from 10- to 20-MeV electrons irradiating homogeneous and heterogeneous phantoms were performed. The detailed benchmarks will appear and appeared in Ref. 11. Fig. 3 and 4 show the results of four sample benchmark simulations.

For the comparison with commercial accelerators in Monte Carlo calculations, we need to model the beam sources of the

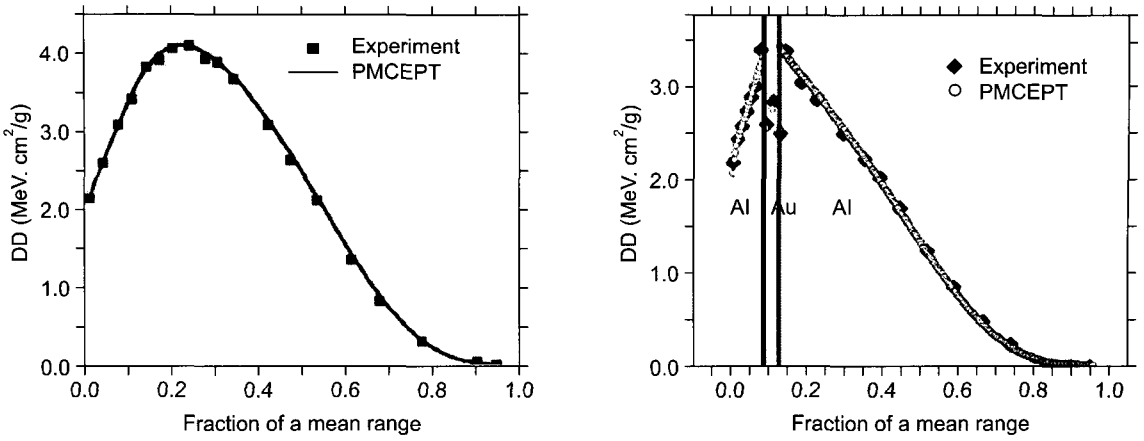

Fig. 3. The benchmark comparisons between PMCEPT and experiment for homogeneous Al (left-hand panel) and heterogeneous Al/Au/Al (right-hand panel) media for low electron energies of 0.521- and 1.0-MeV from left to right, respectively.

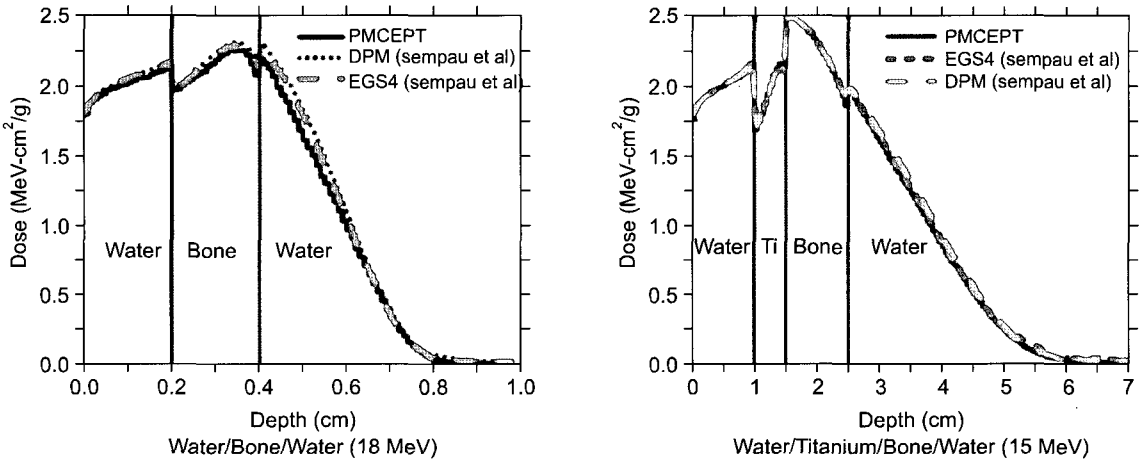

Fig. 4. The benchmark tests between PMCEPT and EGS4 & DPM for two multilayered targets for high electron energies of 18- and 15-MeV from left to right, respectively. The materials used are shown in the figures.

linacs, which require the detailed head information of them. However, this information is usually confidential. A few papers tried to model beam sources for a few commercial accelerators and introduced their head information. We developed template models of beam sources from available open-source head information.[12)]

3. Linear programming for optimization

During radiation therapy, irradiating beams pass through a patient, killing both carcinoma and normal cells. Therefore, the radiation treatment must be carefully planned to deliver a clinically prescribed dose to the target containing carcinoma cells, while nearby organs and tissues are spared. Currently, the intensity modulated radiation therapy (IMRT) is considered to be the most effective radiation therapy for various shapes of carcinoma. Fig. 5 shows the schematic picture of a well optimized radiation therapy.

In IMRT, the patient is irradiated from several beams, each of which is decomposed into hundreds of small beamlets. The intensity of each beamlet can be controlled individually by controlling the beam-on-time. There are a number of potential linear programming approaches to radiation therapy dose optimization. Fig. 6 shows how to formulate the set of constraints represented by the matrix A (p,i,j). We have studied various forms of objective functions, $f(x)$, including linear, mixed integer, and nonlinear functions.[4,13)] The size of A is

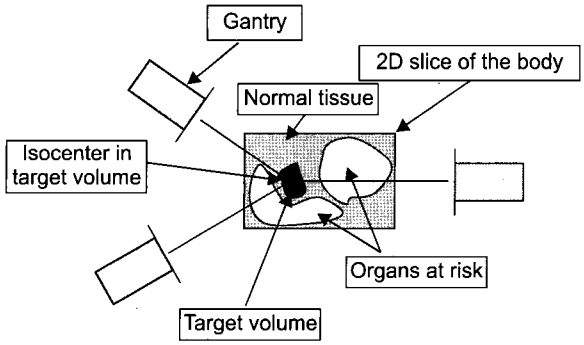

Fig. 5. The schematic picture of a well optimized radiation therapy in two-dimensions. Two organs at risk and one target volume are shown. Three beams (different directions) around one isocenter provide prescribed doses for target, organs at risk, and normal tissues.

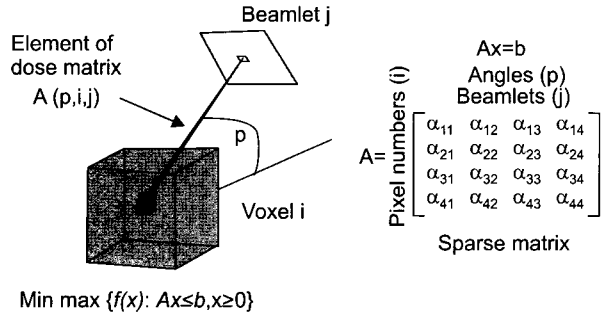

Fig. 6. The schematic picture of IMRT optimization. The matrix A(p, i, j) is calculated by the Monte Carlo simulation. The objective function, f(x) can be chosen from any form of linear, integer, or quadratic functions.

enormous because A has three variables, p the number of gantry angles, i the number of voxels, and j the number of beamlets. One way to reduce the size of A is to choose small number of voxels in the simulation. But the choice of insufficient number of voxels can cause the homogeneity problem. Frequently, we may observe the hot spots that may have critical effects to the organs at risk. We have studied dominating set theory[14] to solve these difficulties. Dominating set theory is a mathematical theory to find the smallest set of positions that dominates the whole area or space.

4. Scientific data visualization

There are many known data visualization tools for simpler tasks such as MATLAB. For complicated or routine works, it might be a good idea to invest some time initially, developing

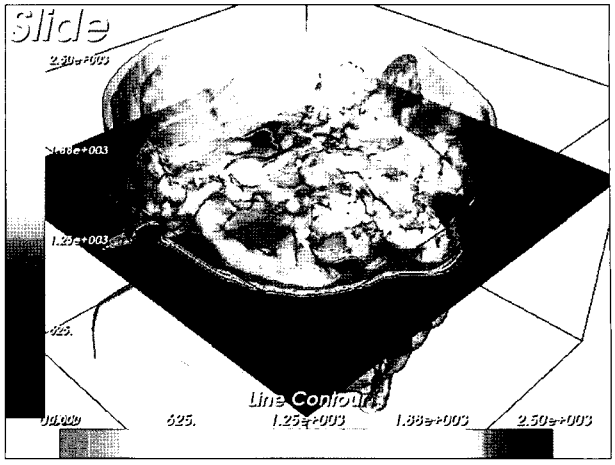

Fig. 7 Sample image picture of 3D image mixed with 2D slice and iso-density curves.

a processing and visualization program with the help of software libraries which provide one to write computer programs to access their interfaces and facilities.

The Visualization Toolkit (VTK) and Insight Toolkit (ITK)[15] are two example free software libraries written by C++. We are developing customized visualization software with many convenient functions for image data such as iso-density curves, iso-dose curves, gray scale and colored image slice pictures, and 3D visualization of image data with transparency. Fig. 7 shows a sample image picture mixed with 3D, 2D, and iso-curves.

A convenient use of visualization tool helps finding the voxel number and associated CT number at the specific point of image (visual automatic CT segmentation). This capability may be useful to choose the minimum number of voxels for optimization simulations.

CONCLUSIONS

We are developing two different client-server customized cancer radiation therapy planning simulation systems based on parallel Monte Carlo calculation and linear programming optimization. The web-based client-server system was introduced in the 48th AAPM meeting.[16] The following critical technologies involved in IMRT, IGRT, and 4D RT planning simulation system, are being developed. (1) Each patient CT data (Hounsfield value) is converted to the physico-chemical

target data which includes density and atomic weight components for each voxel. (2) Each pixel of the CT is characterized if it is in the target volume, or organs at risk, or normal tissues using auto-segmentation technique. (3) The beam head is modeled using MC code or separate beam models. (4) The dose matrix is calculated for each voxel for a given gantry angle and beamlets. (5) To optimize the gantry angle and beamlet weight, simplex, integer, mixed-integer, and quadratic cost or target function is studied. (6) To maintain dose homogeneity while reducing dose matrix size, dominating set theory is used. (7) 2- and 3-D scientific data visualization technique is developed. This will provide high quality normalized services to many hospitals regardless of their locations with a reasonably low cost.

APPENDIX

1. The summary of equations for converting CT number into tissue parameters

Since the detailed description of equations is given in Ref. 5, a brief summary will be described here.

The CT Hounsfield unit H is defined,

$$H = (\frac{\bar{\mu}}{\bar{\mu}_{H_2O}} - 1)1000, \quad \text{(a1)}$$

where $\bar{\mu}$ is the average linear attenuation coefficient according to a location independent spectral function $\hat{f}(E)$ (E: energy). If we assume that the reconstructed attenuation values of the different tissues are completely independent of the location of the tissues, the attenuation values can be considered as mean values ($\bar{\mu}$) according to a location independent spectral function:

$$\bar{\mu} = \frac{\int \hat{f}(E)\mu(E)dE}{\int \hat{f}(E)dE}, \quad \text{(a2)}$$

where the linear attenuation value μ is given by

$$\mu(E) = \rho N_A \sum_{i=1}^{n} (\frac{w_i}{A_i}\sigma_i(E)), \quad \text{(a3)}$$

with ρ(g cm^{-3}) the mass density, N_A (mol^{-1}) the Avogadro constant, i the element index, W_i the elemental weight, σ_i (barn/atom) the total cross section, A_i (g mol^{-1}) the atomic mass. However, $\hat{f}(E)$ is not known for a CT scanner, direct conversion using the above procedure is not possible even though we have tabulated values for the cross section σ. Therefore, we approximate the cross section in a polynomial type of atomic number Z_i,

$$\sigma_i(E) = Z_i K^{KN}(E) + Z_i^{2.86} K^{sca}(E) + Z_i^{4.62} K^{ph}(E), \quad \text{(a4)}$$

where K^{KN} is the Klein-Nishina coefficient. In this equation, we assumed that the attenuation of x-rays with a typical mean energy of 50~100 keV is determined by three different physical processes: photoelectric absorption (ph), Compton process (KN), and Rayleigh process (sca). Then, the mean value of the attenuation coefficient is,

$$\bar{\mu}(E) = \rho N_A \sum_{i=1}^{n} (\frac{w_i}{A_i} Z_i \bar{K}^{KN}(E) + Z_i^{2.86} \bar{K}^{sca}(E) + Z_i^{4.62} \bar{K}^{ph}(E)) \quad \text{(a5)}$$

The coefficients ($\bar{K}^{KN}$, $\bar{K}^{sca}$, $\bar{K}^{ph}$) are mean values with respect to the spectral function $\hat{f}(E)$.

For the attenuation relative to water we get

$$\frac{\bar{\mu}}{\bar{\mu}_{H_2O}} = \frac{\rho}{\rho_{H_2O}} \frac{\sum_{i=1}^{n} (w_i/A_i)(Z_i + Z_i^{2.86}k_1 + Z_i^{4.62}k_2)}{(w_H/A_H)(1+k_1+k_2) + (w_O/A_O)(8+8^{2.86}k_1+8^{4.62}k_2)} \quad \text{(a6)}$$

$$\text{with} \quad k_1 \equiv \frac{\bar{K}^{sca}}{\bar{K}^{KN}} \quad \text{and} \quad k_2 \equiv \frac{\bar{K}^{ph}}{\bar{K}^{KN}} \quad \text{(a7)}$$

The values of (k_1, k_2) are dependent on the CT scanner and are determined experimentally.

2. The PMCEPT code: A brief overview of the physical and computational models

The PMCEPT code embodies physical reality by following the trajectory with particles being created according to the distribution described by the source. Particles travel to a collision distance determined by a probability distribution that depends on the total cross-section and then scatter into another energy and/or direction according to the corresponding differential cross section. New secondary particles can be born when the primary particle interacts with the medium. These secondary particles are also transported in the same way. This procedure continues until all particles are absorbed or leave the region under consideration.

Essentially, there are twelve possible processes by which the electromagnetic field of a photon may interact with matter. Among them, four interactions, Compton scattering, pair production, photoelectric absorption, and Rayleigh scattering, are the most important. The rest are negligible processes. The PMCEPT code includes these four primary scattering processes for photon interactions with matter. A charged particle moving

through a medium interacts with it basically in three different ways depending on its kinetic energy and the distance of the particle from the atom with which it interacts: (1) by collision with an atom as a whole, (2) by collision with an electron, and (3) by radiative processes (bremsstrahlung). When the distance of closest approach to the atom is large compared with the atomic dimensions, the moving charged particle interacts with the atom as a whole. The Coulomb force is the primary interaction force, resulting in an excitation or ionization of the atom. This distance encounter, sometimes called a soft collision, is described by either single or multiple elastic scattering.[17-22] A single elastic scattering simulation is an event-by-event simulation of electron transport, which is often not possible due to limitations in computing power because of the large number of interactions with surrounding matter. To overcome this difficulty, Berger[23] developed the condensed history technique, called multiple scattering. This idea is equivalent to solving a diffusion problem. Lewis[17] showed the exact solution by solving integro-differential diffusion equation, which led to expressions for the various moments of the spatial and the angular distributions. The multiple scattering algorithm implemented in the PMCEPT code solves the Lewis equations. If the distance of closest approach to the atom is of the order of the atomic dimensions, the moving charged particle interacts with one of the atomic electrons. This interaction results in the ejection of an electron from the atom with considerable energy: Moller scattering for an electron and Bhabha scattering for a positron. In general, the energy transmitted to the secondary electron is large compared to the binding energy, and the process can be treated as a free electron collision, but the intrinsic magnetic moment (spin)[24] of the charged particle must be taken into account in the collision probability. In particular, when the two particles are identical, exchange phenomena occur and become important if the minimum distance of approach is of the order of the deBroglie wavelength. When the high-energy charged particle pierces the electron clouds of an atom and approaches close to the nucleus of the atom, the electric field of the nucleus strongly deflects the particle. This deflection process results in radiative energy loss and the emitted radiation (bremsstrahlung)[25-28] covers the entire energy spectrum up to the maximum kinetic energy of the charged particle. The basic formulae for this process are taken from the review article by Koch and Motz.[25] Butcher and Messel's idea[29] for mixing the cross sections for sampling of the secondary spectra is added. When a charged particle travels through a condensed medium, it loses energy both via the hard collisions explained above and via an average stopping power effect (soft-collision term).[18,30] The collision and radiative stopping power and the range tables are taken from the National Institute of Science and Technology (NIST) website and the widely accepted Internal Commission of Radiation Units and Measurements (ICRU) tabulations.[31,32] To describe the soft-collision energy loss process, we use the continuous-slowing-down-approximation (CSDA), which is assumed to be equal to the stopping power. In this approximation, fluctuations in the energy loss are disregarded, and the energy of the particle is taken to be a deterministic function of the path-length traveled.

Since the Monte Carlo method is inherently parallel due to the independent nature of particle transport, the use of parallel processing for the Monte Carlo simulation offers an attractive approach toward improving the overall computational time. Parallelization of the Monte Carlo code is a straight forward approach and is based on distributing the job (number of histories) among different processors that work independently of each other in parallel. Usually, the processors are divided into two classes in this parallel computing. One is the master processor, and the others are slave processors. Because all the processors share the same target information, they do not need to communicate during the calculations, but their final result is passed to the master node when they finish their calculations. Thus, the parallel efficiency is directly proportional to the number of processors employed. The most important problem for this type of parallel processing is to generate a completely independent series of random numbers on each processor. This problem was discussed in detail in Ref. 3. The structure of this parallel program is the single program multiple data (SPMD) structure in the sense that the master processor controls the entire job as well as its own computing job. The standard message passing interface (MPI)[6] was implemented for the communications between processors. The MPI is a standards-based message passing library for a set of processing elements, typically with distributed memory. It is also one of the most popular interfaces for parallelizing existing serial applications.

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네트워크기술을 이용한 서버-클라이언트 원격의료기술은 특히 의료시설이 낙후된 지방도시의 의료기관에 질 높은 의료서비스를 제공할 수 있는 기술이다. 이러한 기술은 중앙 집중 방식으로 진단과 검사용으로 사용되는 대형 컴퓨터 하드웨어와 소프트웨어를 매우 효율적이고 경제적으로 관리할 수 있게 하게 때문에 궁극적으로는 의료수가를 감소시키는데도 기여할 것이다. 각 환자에 대해 환자 맞춤형 방사선 치료계획은 매우 효율적인 암 치료를 가능하게 하기 때문에 환자와 의사 모두에게 매우 유익한 방법이다. 치료계획전문가들은 환자에게 너무 적은 선량을 주면 암이 계속 재발할 확률이 높고 너무 많은 선량을 주면 환자를 다치게 할 수도 있다는 것을 잘 이해한다. 최고의 해법은 가장 정확한 선량을 주는 것인데 이것은 각 환자의 CT 자료를 기반으로 정확한 선량계획 시뮬레이션 시스템을 사용하는 것이다. 우리는 네트워크 기반과 웹 기반을 이용한 환자 맞춤형 치료계획 시뮬레이션 시스템개발을 위해 관련된 4가지 컴퓨터 프로그램을 개발하고 있다; 환자의 CT 자료를 이용하여 각 환자의 표적 자료를 만드는 프로그램, 이 표적자료를 바탕으로 방사선 선량 시뮬레이션을 하는 병렬 몬테카를로 프로그램, 선량주사변수들을 최적화 시키는 프로그램, 그리고 계산결과를 시각화하는 프로그램들이다. 모든 소프트웨어는 약 100~200개의 개인컴퓨터로 구성된 클러스터에서 병렬모드로 운영이 된다. 이와 같이 방대한 하드웨어와 소프트웨어의 효과적인 관리를 각 병원에 맡기는 것은 효율적이지 못하기 때문에 이를 중앙에서 관리하면서 각 병원에서는 네트워크나 웹을 통하여 마치 모든 것이 자기 병원에 있는 것과 같이 편리하게 쓸 수 있게 하는 시스템으로 의사와의 계속적인 의사소통은 클라이언트-서버 시스템의 메신저 기능을 이용한다.

중심단어: 원격의료, 클라이언트-서버시스템, 환자맞춤식 치료계획 시뮬레이션 시스템, PMCEPT 코드, 베어울프 PC 클러스터