THE STATE OF MONTE CARLO CALCULATIONS IN RADIATION TREATMENT PLANNING

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Received December 4th, 2001; received in a revised form April 5th, 2002; accepted April 11th, 2002

INTRODUCTION

The present paper is an overview of the current status and development of Monte Carlo (MC) techniques in radiation treatment planning (RTP). It will address the following questions: Why does one want to use MC instead of or in addition to the current advanced treatment planning systems; what accuracy may be achieved in relation to the time effort; is or will MC be feasible in the routine treatment planning process and what are the approximations and tricks applied to make is feasible. In short: how far does MC represent reality, what is the place of MC in the most advanced treatment techniques such as conformal therapy and IMRT? It will be seen that general purpose of MC codes, such as EGS4 [1], have to be adjusted to the special needs of RTP in order to make them practical. From such adaptations originate the prominent MC methods in RTP known as VMC [1] (V for voxel) and XVMC [2], (X for x-ray), MCV [3], (V for Vista), MCDOSE [4],, MMC [5], (first M for macro) or the PEREGRINE effort, [6], the main features of which will be shown.

WHY MONTE CARLO?

The accuracy of currently available dose computation models for planning of radiation treatments is limited since empirical approximation has to be used for the descriptions of radiation source models, radiation beam collimation and characteristics and interaction of radiation with matter, and patient (or phantom) or tissue inhomogeneities. The discrepancies of TPS results compared with real dose distributions may be clinically significant in many cases. The “real” dose distribution is understood as the one generated in actual irradiations of patients or phantoms and could be closely approximated either by high precision measurements in phantoms or with general purpose MC codes run with large numbers of particle histories (in patients and phantoms). If such discrepancies could be revealed by accurate predictions of dose they could be remedied using different treatment techniques such as the use of different energies, beam arrangements, and intensity modulation.

Theoretically, MC has the potential of delivering true dose distributions provided that high accuracy practical reach and affordable with Monte Carlo simulations of radiation transport. MC inherently provides a universal accuracy, i. e. all materials, modalities, anatomic geometries, devices may be accounted for. MC eliminates the need for laborious trial and error parameterization and refinement of models, which is common in conventional planning systems. The additional advantage is reduction in time and the amount of measured dose distribution data required for commissioning and validation. Thus MC may represent a simpler planning method.

Keall [2] has pointed out further advantages of MC treatment planning: the possibility of directly predicting monitor units accompanied by a reduction in the probability of human mistakes; improvement in the consistency of inter-institutional results as well as in the quality of dose response data; and accurate estimation of quantities difficult or impossible to measure. MC makes it also possible to develop/test “virtual” devices and to solve problems that do not have analytical solutions.

DO MONTE CARLO RESULTS REPRESENT REALITY?

MC calculation methods aim at realistically simulating the actual interaction processes involved in particles transported through matter. Single particles representing real particles are simulated. During a real treatment a 2 Gy fraction requires about 10^{16} electrons incident upon the target and about 10^{14} photons impinging on the patient. However, due to present limitations in computer technology particles “only” in the order of 100 million are used in patient simulations in order to achieve an acceptable compromise between sufficient accuracy and time needed for the planning process.

As will be shown below, in order to make MC RTP feasible some simplifications and approximations are introduced into the simulation algorithm, mostly variance reduction techniques, that remove MC results further from reality. Basically what is being applied is the formalism of the simulation process which allows us to still call it a MC technique. However, computer technology is still developing rapidly and with greater speed of calculation an approximation representation of reality will gradually be possible. Even now the available MC based TPSs have been shown to be superior to conventional planning algorithms in terms of accuracy [3,4]. Examples have been shown during the authors oral presentation.

ACCURACY AND “NOISE”

The cost of the above mentioned compromise is that an increasing number of particles transported will increase computing time linearly but will improve statistics only by the square root of the number of particles. In other words: Since Monte Carlo simulation results suffer from ‘noise’ and this noise decreases with the number of particles, accuracy decreeases as the square root of time.

Noise as well as systematic errors have an influence on dose-volume histograms (DVHs). It has been shown that

- the dose to organs at risk is less sensitive to noise than a target dose,
- systematic errors affect biological indexes (TCP and NTCP) to a significantly higher extent than random noise,
- a random noise level of $\leq 2\%$ does not significantly affect isodose lines, DVHs or biological indexes [2].

One characteristic of MC planning is that the number of beams simulated does not necessarily increase the number of histories to be simulated proportionately. For example the total number of particles depositing energy in the target area, determines the statistical accuracy or noise, rather than the number of histories per beam. However, larger beams mean more histories that reach the same noise level and this increase is proportional to the field area.

ABSORBED DOSE TO WATER AND TIS-SUE

MC simulation delivers results expressed in terms of dose to tissue. Current clinical experience in radiation therapy is, however, based upon the absorbed dose to water computations. A method to convert the dose to the medium to dose to water is therefore desirable. A single correction factor for each material may be used with an added error of $< 1\%$. [2].

MC in IMRT

Intensity-modulated radiotherapy (IMRT) dose calculations are especially sensitive to inaccuracies in the respective dose calculation algorithms. Many IMRT dose calculation systems use fast Pencil Beam (PB) or Superposition/Convolutions (SC) algorithms for computing dose. PB algorithms are however known to be inaccurate in heterogeneous media. SC algorithms are also inaccurate in some instances. Due to the improved accuracy Monte Carlo (MC) algorithms are considered most appropriate in IMRT dosimetry. IMRT DVHs in target and organs at risk show differences in the outcome of the three algorithms mentioned. Table 1 shows the effect of the algorithms on biological response functions.

Tab. 1. Response functions in IMRT dosimetry derived with three algorithms in a laryngeal cancer case treated with parallel-opposed multi-sectional portals of 6MV photons [3].

	Monte Carlo	**Superposition**	**Pencil Beam**
TCP	86%	90%	89%
NTCP			
Spinal Cord	1%	1%	0%
Right Parotid	92%	95%	95%
Left Parotid	2%	3%	0%
Larynx	4%	3%	3%

TCP = Tumor Control Probability

NTCP = Normal Tissue Complication Probability

In summary, the following considerations apply to IMRT dosimetry [2]:

- MC better predicts in-phantom and in-patient dose distributions,
- DVHs differ for MC, PB and SC dose calculation algorithms,
- TCP and NTCP vary with the dose calculation algorithm. Whether not this variation is clinically significant remains to be determined, and
- MC may be required for dose distribution optimization.

In respect to optimization distinct differences between conventional radiotherapy treatment techniques and IMRT techniques may be noted. In conventional plans, a trial-and-error approach with few unmodulated open or wedged beams is employd. An experienced user may be able to estimate the effect of the inherent inaccuracies of the algorithm although it should be advantageous to have an accurate algorithm available. IMRT however, provides automatic optimisation with a few or many highly modulated beams, and it is virtually impossible to assess the effect of inaccuracies of an algorithm on the credibility of the resulting dose distribution. The use of an MC code, therefore, appears to be essential in IMRT planning.

METHODS OF MONTE CARLO SIMULATION IN RADIATION TREATMENT PLANNING

MC dose calculations are commonly performed in a voxel geometry as derived from computed tomography (CT) imaging. The “sharp” tracks of primary and higher generation particles in a medium are spread out into a distribution pattern of voxels in which a certain amount of dose is deposited. Primary and higher order electrons, deposit doses continuously on the scale of the usual voxel size. Their “voxel tracks” are contiguous, whereas with photons it is only the “voxel tracks” of the secondary and higher order electrons that show up with gaps between over the free path length of the photon. This change in the dose deposition pattern from real to voxel tracks is compensated for statistically by a large number of simulated particle tracks or particle “histories”.

The least prestigious application of MC simulation to RTP may be seen in applications of a mere supportive character. For example, dose distribution kernels in materials of various composition as needed in SC or pencil beam algorithms may be determined by MC. Another example may be seen in the determination of monitor units using MC, yet another one in the investigation of the influence of a metallic block in a radiation field, etc. General purpose MC codes such as EGS4 (or EGSnrc) could of course be used more or less directly, and have been used in this way for dose distribution calculations in radiotherapy. This process is, however extremely time consuming and not practical in clinical routine applications. Such use of MC simulations has been and still is of interest in special situations, where a general behavior is to be investigated, or even in a clinical case with a special problem, e. g. a metallic implant in a radiation beam, when no MC TPS is available

(see *Figures 1-3*). The advantages of tailoring a general MC code to the needs of radiotherapy is obvious if one realizes the range of applicability. For example, EGS4 is applicable to materials with elements of atomic numbers up to Z = 100 and an energy range of 1 keV to10^5 MeV. In contrast, in radiotherapy usually only low-Z elements, say up to Z ≈ 20 (excluding, however, medium- and high-Z metal implants), and an energy range of 100 keV to 25 (50) MeV apply.

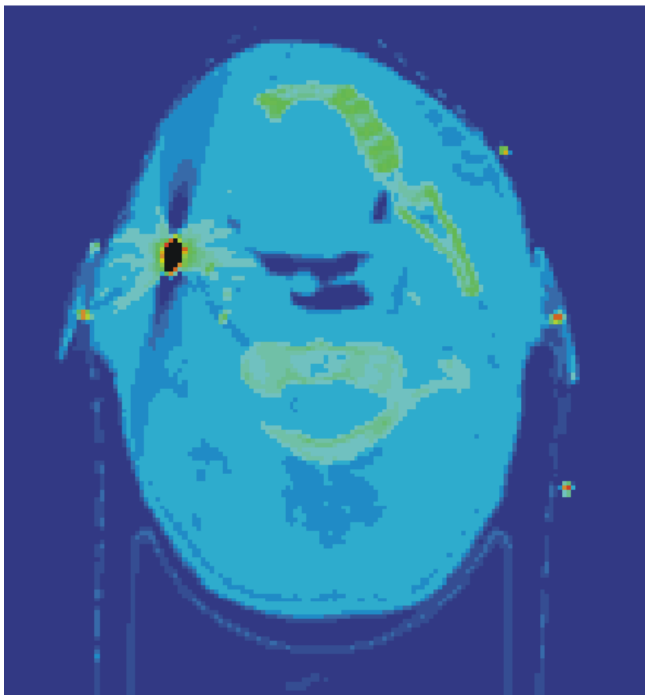

Fig. 1. Radiotherapy planning CT image in the head of an electron therapy patient showing an image artifact from a titanium jaw prosthesis after surgical tumor resection.

Fig. 2. The CT cross-section of Fig. 1. after artifact removal and generation of bulk inhomogeneities.

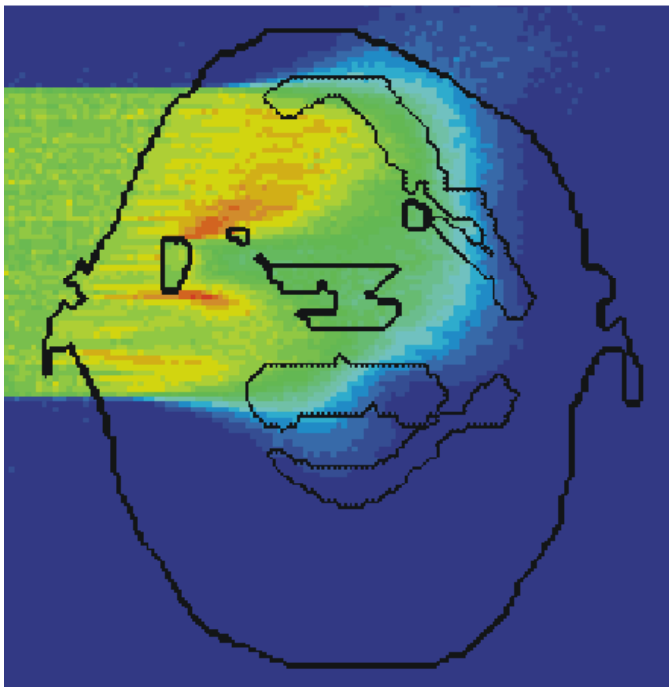

Fig. 3. EGS4-generated dose distribution for the cross section of Fig. 2 for a lateral 10 MeV electron beam, field size 10 x 10 cm^2, showing the impact of the titanium jaw replacement.

Since in MC applications usually millions of particle histories are simulated it is suitable for parallel processing. Consequently, general purpose MC codes as well as specialized codes have been adapted to this type of processing, in some cases with a multiple array of 60 or more PC's.

As mentioned earlier in the introduction, a variety of specialized MC treatment planning codes have been developed for single desk-top computer operating systems with, e.g. systems using Pentium II or higher processor and frequencies from 450 MHz upward, or similar Windows.

SOURCE MODELS

The starting point for MC simulation is the description of a radiation source. Since in a real treatment unit the radiation impinging on a patient does not strictly originate from a point source or target but has instead a rather complex structure modeling of the source of radiation is necessary. Frequently, a combination of a point source for the primary radiation and one or more extended sources (multiple source model = MSM), e. g. with a Gaussian intensity distribution, are applied. Radiation from that source model may then be transported through the treatent head to define a source model of raiaion impinging on the patient. This source model may additionally be transported through patient-dependent beam modifiers during the individual radiotherapy planning.

The BEAM [1] code added to the EGS4 or EGSnrc systems was designed to transport radiation from a point source at the exit window of an accelerator through the construction parts of the treatment head such as flattening filters or scattering foils, beam light mirror, monitor chamber(s), upper and lower x-ray collimators or electron beam shaping cones. In a trial and error procedure of reproducing the measured depth-dose curves and cross-beam profiles the source energy and spectrum may be identified. The BEAM produces a full phase-space file of particles in a chosen plain, e. g. in the cross-beam plane at the lower jaw or cone end or at the patient's or phantoms surface. Such phase-space data may then be used as the source model of the radiation which is subsequently transported through some additional devices or through the patients body. Phase-space-files are of course large and not easy to handle. Simplified source models are therefore desirable. The BEAM also has a useful feature that it can help to determine which part of the radiation originates from which part of the treatment head, and enabling

the development of effective simplified source models. It can also be applied to treatment head design optimization and other tasks.

BENCHMARK TESTS

As we have seen specialized MC planning systems are derived through approximations and simplifications by means of which a general MC code is adapted to the special need of RTP. It is even more necessary than general systems [8] that the system should be tested in all aspects relevant to correct patient treatment. Consequently, a number of benchmark tests have been suggested. A basic test involves the reproduction of depth dose curves in a water phantom or a phantom containing layers of materials with densities other than water. Also cross beam profiles in homogeneous or inhomogeneous arrangements may be reproduced. A test that demonstrates a better representation of reality with MC codes is a single beam distribution in a inhomogeneous patient cross section when only a part of the beam passes through the inhomogeneity. A correct representation of side scatter is especially difficult to achieve with advanced conventional planning algorithms but inherent in a MC simulation.

SPECIALIZED MC TREATMENT PLANNING CODES

In what follows some features of the current, more or less arbitrarily selected special purpose MC TPSs will be discussed in order to give the reader an idea of what is involved in tailoring MC to the radiotherapy problem and what is the present state of the art.

FEATURES OF THE MCDOSE CODE

The MCDOSE is an EGS4 user code for 3D dose calculation suitable for conventional photon and electron beams and IMRT planning and verification. It employs accurate source models and convenient commissioning tools, allows for the simulation of beam modifiers and was implemented with variance reduction techniques. The electron and photon source models are multiple source models derived from MC simulation of measured beam data. The commissioning process is automated. The authors claim an accuracy of < 2% of the maximum dose everywhere in the calculated dose distribution. The agreement with the BEAM/DOSXYZ code and the general EGS4 code as applied to dose calculations as well as with measurements is within 2%. The MCDOSE is 27 times faster than DOSXYZ in a benchmark test suggested by Rogers and Mohan [9]. The CPU-time for a 9-field IMRT plan for pre- and post-optimisation calculation is typically 1-4 h using a 450 MHz PC.

MAIN CHARACTERISTICS OF THE VMC AND XVMC [2]

The VMC is a fast electron transport code which incorporates the simulation of delta electron and bremsstrahlung production. It uses a condensed history technique and continuous boundary crossing. The XVMC is based on the VMC, i. e. electron tracks are handled according to the VMC. Photon radiation is modeled as single scatter MC accounting for photoelectric absorption. Compton scattering and pair production. The accelerator is modeled by extracting parameters from measurements in water (^{w}) and air. Material attenuation coefficients for Compton scatter (^{c}), pair production (^{p}) and photo absorption (^{a}) are approximated by the corresponding water coefficients and an empirical density function

$$\mu_{c(p,a)}(E,\rho) = \mu^{w}_{c(p,a)}(E) \cdot f_{c(p,a)}(\rho)$$

where E is the particle energy, ρ is the density, and f is the density function. For example density function for Compton scattering, is

$$f_c(\rho) = \frac{n_e(\rho)}{n_e^w} \approx \begin{cases} \rho/\rho^w, & \rho \le \rho^w \\ 0.85\rho/\rho^w + 0.15, & \rho \ge \rho^w \end{cases}$$

where n_e is the electron densitiy.

There is not enough space here to explain other sophisticated time saving techniques for primary and higher order electrons and photons, such as electron

history repetition, photon splitting and the so-called Russian Roulette, all involving a repeated use of pre-calculated particle histories at various interaction sites. Also incorporated is a "KERMA approximation" in which electron transport is switched off if the energy of higher order photons is lower than a selected limit K_{cut}. and the energy is deposited along the photon path using the energy absorption coefficient. Instead of uncorrelated random numbers the SOBOL's sequence of quasi-random numbers is applied. Transport parameters such as photon and electron energy cut-off, KERMA cut-off, and maximum electron step size are optimized.

The agreement of the VMC and XVMC codes with the general EGS4 or EGSnrc codes is usually within 2% in water and tissues such as lung or bone. The calculation times using a 450 MHz PentiumII PC for a water phantom with 5mm voxels are for 6 MV and field sizes of 5x5, 10x10 and 20x20 cm² 65, 242 and 928 sec, respectively. For 18 MeV they are about 30% higher.

Future improvements are foreseen in the development of an improved accelerator head model including off-axis softening, head scatter contributions and monitor unit prediction and in beam modifier simulation (wedges, compensators, multi-leaf collimators), as well as in IMRT optimisation.

CONCLUSIONS

MC-based RTP is clinically feasible at present because of the hardware-to-price performance ratio doubling every 18 months [5] and because of the availability of specialized radiotherapy MC codes such as those mentioned, and due to accurate source models available. MC codes are also suitable for parallel processing with parallel hardware or in distributed computing.

In terms of accuracy, MC methods are superior to traditional RTP dose calculation methods.

In terms of speed, fast specialized MC RTP codes present themselves as a real alternative to traditional RTP dose calculation methods, especially in electron beam treatment planning where they will soon become the method of choice. They will be an indispensable tool for photon beam RTP, especially in IMRT.

Acknowledgement: I am grateful to the following persons for providing me with a wealth of frames and ideas: A. F. Bielajew, Ann Arbor, MI, USA, M. Fipppel, Tübingen, Germany, I. Kawrakow, Ottawa, Canada, C.-M. C. Ma, Stanford, CA, USA, R. Mohan with P. Keall and J. V. Sievers, Norfolk, VA, USA, H. C. Neuenschwander, Berne, Switzerland and the PEREGRINE-Group of Lawrence Livermore National Laboratories, Stanford, CA, USA. The case represented in Figures 1-3 have been produced by my co-workers Marc Lauterbach and Joerg Lehmann.

REFERENCES

The authors of the following references, except reference 7, provided frames through the internet to the present author. Since this presentation actually refers to the frames provided the citations below are intended to give credit to these authors and some of their recent work. They may also serve as a guide to further related sources.

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