Monte Carlo Based Dose Estimation in Intraoperative Radiotherapy

Pedro Guerra, Wilfredo González, Maria J. Ledesma-Carbayo, Jacobo Cal-González, Elena Herranz, Jose M. Udias, Antonio Lallena and Andrés Santos

***Abstract*– This work presents the application of an updated version of an existing Monte Carlo (MC) code for photon and electron transport to intraoperative radiotherapy (IORT) dose planning. The code has been optimized to provide an estimation of the dose delivered by the linear accelerator (LINAC) in very short time and has been incorporated as a submodule of an advanced IORT planning system under development, which will enable far more accurate pre-planning as well as the intraoperative treatment modifications considering actual information taken at the operating room (OR) during surgery, for instance with a mobile C-arm.**

I. Introduction

Intra Operative Radiation Therapy (IORT) refers to the application of high energy radiation during a surgical intervention, after the resection of a neoplastic mass. Conventionally, IORT uses a single direct irradiation on the residual tumor or tumor bed area while sparing normal surrounding tissue. Besides the advantage of direct access to the irradiation site, IORT may also permit to preserve healthy neighboring organs either by dislodging them from the bulk of the radiation field or by interposing shielding between them and the target area. These characteristics enable achieving better radiotherapeutic conditions than the traditional techniques.

However, unlike conventional radiotherapy, IORT treatment planning has been traditionally highly unsophisticated, exclusively limited to the consultation of graphs and charts containing the isodose curves measured under standard conditions. Recently Radiance (GMV, Tres Cantos 28760 Madrid, Spain), is being designed to cover this niche. The surgical planner needs to cover three phases:

- Pre-operative: simulation of the surgical conditions, planning of the applicator position and calculation of the dose based on a pre-operative CT scan of the patient.
- Intra-operative: verification of the applicator position and dose recalculation with the actual patient location, based on an intraoperative imaging scan.
- Post-operative: recording and reporting of the actual delivered dose.

Traditionally dose planners have implemented approximated methods, such as pencil beam [1] or collapsed cone [2], being the golden reference for validation the output of realistic Monte Carlo (MC) Simulators, such as EGS [3-5] or MCMP [6], which are considered precise but are inherently computationally demanding, with computation times too long for intra-operative clinical environments.

Approximated methods show limitations in dose computation for non-homogeneous materials; for instance, pencil beam models rely on an analytical description of the total energy distribution released in a semi-infinite volume, usually derived from a Monte Carlo simulation of photons/electrons impinging on a semi-infinite slab of material. This approximation could cause errors when dealing with small irradiated volumes, limited in the lateral and/or forward direction. The overestimation of phantom scatter in these conFig.tions yields an overestimation of the calculated dose. The modeling of patient curvature in a pencil beam model can also introduce errors when dealing with small volumes [7].

In order to overcome these limitations, several authors have worked on simplifications of the physics to speed up simulation time, such as the voxel MC algorithm VMC++ [8], MCDOSE [9], Peregrine [10] and DPM [11]. These algorithms calculate 3D dose distributions for standard radiotherapy conditions in a time frame that is typically one to two orders of magnitude shorter than conventional Monte Carlo codes[12].

These developments have motivated some companies, such as CMS, Elekta, Brainlab (XVMC), North American Scientific (Peregrine), Nucletron (VMC++) or Varian (MMC and VMC++), to integrate MC simulation in their dose planners.

Compared to external radiotherapy, IORT imposes a hard constrain on the available time for computation, as the patient lies on the operating table waiting for the dose to be delivered. This work deals with the development of a MC-based dose tool that can be used in any of the planning phases of IORT. The goal is being able of providing a accurate dose estimation,

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i.e. with uncertainty 2σ/DMAX <2%, in a reasonable time, around couple of minutes. This tool optimizes an existing MC algorithm for its execution in a multiprocessor platform and integrates all required steps for dose computation in IORT.

II. Material and Methods

A. Imaging Protocol

The MC kernel is meant to replace the existing dosimetry computation engine, based on an adapted version of the pencil beam algorithm [13] used in external radiotherapy.

The IORT dose planner takes as input either preoperative or intraoperative CT volumes. These volumes are linked, together with the LINAC applicator and the patient, to the OR physical coordinate system via 3D position trackers. This way, at every time point it is possible investigate approximation approaches with the applicator for dose delivery during the pre-operative phase, with haptic interfaces that render a force response to the user and thus avoid penetrating rigid structures and alternatively locate and record the actual position of the applicator within the body during the intraoperative for dose replanning and reporting in the post-operative phase. Fig.1 shows cervix irradiation on a mini-pig during a surgery training session. The left picture shows the surgical field as seen in the OR; the right picture represents the same situation in the planning tool.

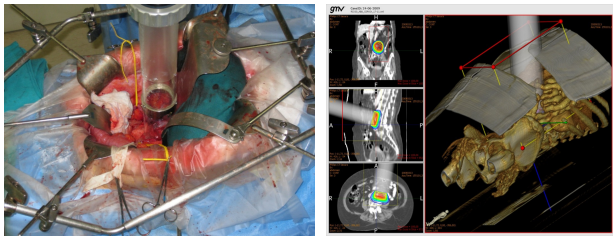


Fig. 1: (Left) Surgical field during an operation. (Right) Rendered virtual representation of the applicator location within the patient's body.

Based on the CT images, the planning tool infers tissue properties at the operating energy of the LINAC combining the stoichiometric methods described in [14], and image segmentation for non-tissues such as bed or bolus. These properties, together with protection location, applicator position and LINAC model are required to estimate the dose in the region of interest.

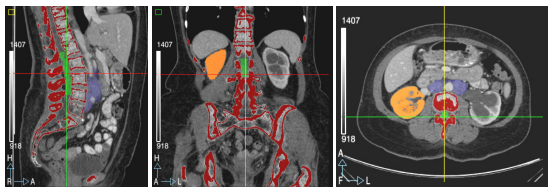


Fig. 2: Segmention with tissues to be removed (light orange) and tissue at risk (red)

B. Monte Carlo Code

Accurate dose estimation requires a correct modeling of the particle source, i.e. the mobile linear accelerator, and the particle-matter interaction process.

The accelerator head model includes electron beam energy spectrum as well the effects primary/secondary collimator, flattening filter, monitoring chambers and electron applicator. In the model of the head each LINAC component is considered as sub-source, which has its own energy and fluence distributions [15]. The model head can be modeled either offline, where particles are stored in a phase-space dataset, or online. Currently, an efficient and accurate head model is under development to compute machine specific phase space datasets, which will be stored using the IAEA phase space data format [16].

Radiation-matter is simulated with an optimized version of the DPM code, which has been ported to C/C++, profiled for maximum speed on multicore computers running either Linux or Windows. DPM has been selected as reference code for its accuracy, computational efficiency and open licensing terms. It was developed explicitly for fast simulation of dose deposition in external electron beam radiotherapy.

As many other Monte Carlo codes do, DPM models electron transport above the keV range by dividing each electron track collisions into a series of tracks which model large numbers of collisions in a single "condensed history" [17] straight-line step. At the terminus of each transport step, the aggregate energy loss and angular deflection over the thousands of collisions making up the step is sampled from distribution functions modeling the physical processes, and the particle is transported as though it underwent a single collision. Clearly, the speed of a Monte Carlo electron transport program will be determined by the number of condensed history steps it must take, which in turn will be determined by the number of collisions it can accurately model in each of the steps.

C. Code Verification and Validation

DPM's dose accuracy for electron beam calculations in heterogeneous media has already been investigated through a comprehensive set of measurements and calculations [18]. The correctness of port to C has been verified by checking long execution traces of the new code against the reference Fortran code, where no intermediate variable showed a difference higher than the least significant digit.

Accuracy of calculations is verified against the EGS4/PRESTA code system. This provides a nominal way of demonstrating the accuracy of the in-phantom portion of the calculations and this well known code has been extensively benchmarked against experiment[19]. The phantom is 1 to 2 cm is water, 2 to 3 cm is aluminum, 3 to 6 cm is lung material and 6 to 30 cm is water. The incident beam is to be a monoenergetic 6 MeV or 20 MeV spectrum from a point source at 100 cm SSD and collimated to 10 cm x 10 cm at the phantom surface.

Accuracy of calculations is verified against experimental results in a water phantom. Dose delivered by a Primus Mid

(Siemens, Erlangen Germany) accelerator in a MP3 water phantom tank (PTW, Freiburg, Germany) is recorded with a Roos chamber and a Tandem electrometer (PTW, Freiburg, Germany). The MC code is executed on a 2800 MHz dual Quad-Core AMD Opteron (Advanced Micro Devices Sunnyvale CA, USA).

In addition to previous phantoms, execution performance for electrons is also tested with a water phantom is 30.5 cmx39.5 cmx30 cm deep and filled with 5 mm3 voxels and using the same incident beam as before.

III. Results

DPM Depth-dose curves for electrons are compared against EGS4/PRESTA in Fig. 3. Likewise Fig. 4 shows the simulated and experimental depth-dose curve and transverse profiles for different depths in a water phantom with a Primus Mid (Siemens, Erlangen, Germany) accelerator, whose beam spectra for electrons and gamma is shown in Fig. 5.

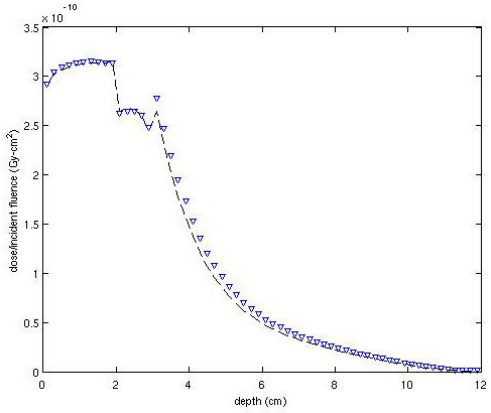
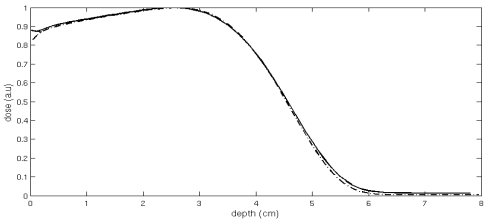


Fig. 3: DPM (blue triangles) vs EGS (black dashed line) for 18 MeV electrons (right).

As expected, depth-dose curves exactly match values predicted by EGS4/PRESTA and values measured in a water phantom.

Regarding simulation speed, seven concurrent threads are executed on a Dual Xenon Quad Core (Intel, Santa Clara, USA) running Ubuntu, with results summarized in Table I. These data are very competitive with an alternative massively parallel implementation on GPU [20].



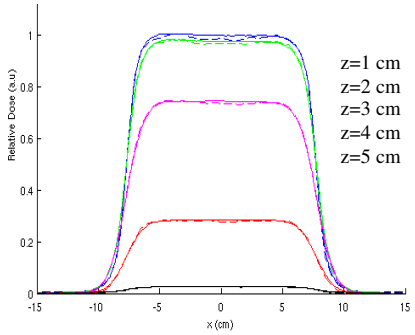


Fig. 4: DPM (dashed) vs experimental data (solid). Longitudinal (top) and Transverse (bottom) profile.

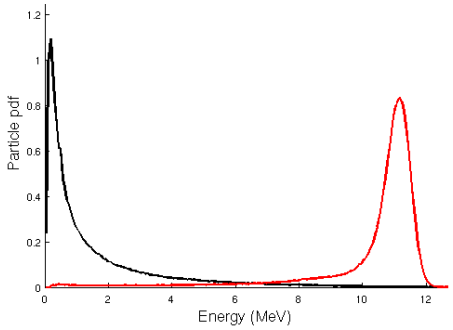


Fig. 5 : Energy spectrum for electrons (red) and photons (black) used to model Primus Mid (Siemens, Erlangen Germany) accelerator.

Table I: Simulation speed with the described phantoms with 7 threads on a Dual Xenon Quad Core 48 RAM.

Phantom	**Energy (Mev)**	**Accuracy**	**Speed (kparticles/s)**
Water-Al-Lung-Water	18	0.25 %	145
Water-Al-Lung-Water	6	0.25 %	363
Experimental Water	11	0.6 %	321
Water	20	2%	215
Water-Lung-Water [20]	20	0.6%	125
Water-Bone-Water [20]	20	0.6%	98

Computation speed is verified with a realistic scenario. The actual number of histories to be simulated depends on the desired uncertainty, volume size and applicator's cross section. Fig. 6 shows a torso CT volume that is converted into densities and tissues. The patient's colon is irradiation with a 6 MeV Siemens Oncor, with source model taken from [16], and a 4 cm radius applicator is simulated, taking 136 s (175 khists/s) to meet the 2% uncertainty requirement. Further code speed up through variance reduction is under investigation.

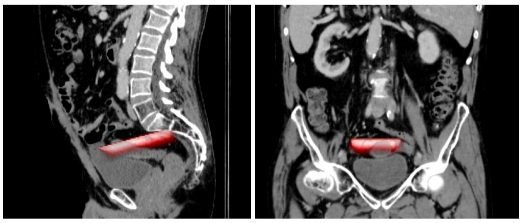


Fig. 6: Dose estimation (red) at the colon for a 6 MeV Siemens Oncor (Siemens Oncology Care Systems, Concord, USA) electron source after simulating 7M histories on a Dual Xenon Quad Core.

The ported code has also been compiled under Windows 7 (Microsoft, Redmond, USA), to test the feasibility of using this simulator as part of the Radiance (GMV, Tres Cantos 28760 Madrid, Spain) dose planner. It takes around 7 minutes (60kelectrons/s) to solve the delivered dose on the colon on a Intel Quad-core I7, as shown in Fig. 7.

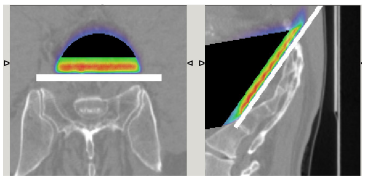


Fig. 7: Dose estimation at the colon for a 6 MeV Siemens Oncor (Siemens Oncology Care Systems, Concord, USA) electron source after simulating 25M histories on a Quad core I7.

IV. Conclusions

A MC simulator tool valid for IORT is being developed after recoding and optimizing an existing dose code. Currently computation speed per core used is above 15,000 histories per second in scenarios of clinical interest. This Fig. shows the feasibility of using Monte Carlo for dose estimation with IORT with response time lower than five minutes. Current research focuses on analyzing the limitations of the simplified physics under DPM and on comparing current code against existing pencil beam solution on specific clinical problems.

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