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ARTICLE

A GPU-Based Track-Repeating Algorithm for Dose Calculation for Photon Radiotherapy

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An essential ingredient in radiotherapy is the calculation of the dose to be delivered to the patient. Analytical algorithms are commonly used for such a task, however their accuracy is not always satisfactory. Monte Carlo techniques provide higher accuracy, but they often require large computational times. Track-repeating algorithms, for example the Fast Dose Calculator, have shown promise for achieving the accuracy of the Monte Carlo approach for proton radiotherapy dose calculations, while considerably reducing the calculation time. We report on the implementation of the Fast Dose Calculator for photon dose calculations on a GPU architecture. As in the proton case, this implementation reproduces the full Monte Carlo dose calculations within 2%, while achieving a statistical uncertainty of 2% in less than one minute utilizing one single GPU card.

KEYWORDS: photon , track-repeating, GPU, dose calculation, Monte Carlo, fast dose calculator, radiotherapy

I. Introduction

Cancer affects approximately one in three women and one in two men. Fortunately, the cure rates for many cancers have been increasing in recent decades, and there are now more than 8 million cancer survivors in the United States alone. There are three major strategies used to treat cancer: surgery, chemotherapy, and radiotherapy. Frequently these modalities are combined to increase tumor control or to reduce treatment side effects. In radiotherapy, there is wide acceptance of the view that considerable benefits could be obtained with a quality increase of treatment plans by reducing the radiation doses to healthy tissues. An essential component for the quality of a treatment plan and tumor response is the accuracy of dose calculations.[1)] The clinical advantages of more accurate dose calculations (i.e., how the treatment plans with higher quality dose calculations will impact tumor recurrence, local control, and normal tissue complications) has not been fully quantified and requires further investigation. Nevertheless evidence exists that dose differences on the order of 7% are clinically detectable.[2)] Moreover, several studies have shown that 5% changes in dose can result in 10%-20% changes in tumor control probability or up to 20-30% changes in normal tissue complication probabilities.[3–5)] Dose distributions for photon or electron radiotherapy for clinical applications are commonly calculated with analytical methods. Such methods are fast but not always accurate enough, especially for highly inhomogeneous media, like the thorax. Monte Carlo techniques, if used appropriately, are known to provide a higher level of accuracy. However the computational times are considerably higher than those of analytical calculations. The availability of always more powerful computers has allowed for the use of a few Monte Carlo codes for clinical applications. However they still need calculation times of the order of minutes on a single CPU to obtain the dose distributions for a typical treatment plan.[6)] Track repeating algorithms applied to proton therapy[7–9)] have been reported in the literature. They use track histories pre-calculated in water with a full Monte Carlo code to quickly estimate dose distribution in heterogeneous media. They have shown important speed gains with respect to traditional Monte Carlo codes. Moreover a track-repeating algorithm for protons, the Fast Dose Calculator (FDC) has been implemented on a GPU based architecture, which reduced calculation times by more than an order of magnitude.[10)] Calculations times can be decreased by performing the computation in a parallel on systems with multiple central processor units (CPUs) or graphics process units (GPUs). Dose calculations carried in parallel on a large number CPUs has also been reported.[11)] GPU clusters are more affordable and require lighter maintenance than traditional CPU clusters. GPU algorithms have been used for dose calculations by Hissoiny *et al.*[12)] and Jacques *et al.*,[13)] who implemented superposition convolution algorithms for dose calculations on GPUs. A GPU-based Monte Carlo code for coupled electron-photon transport was implemented by Jia *et al.*,[14)] who reported a calculation time reduction up to a factor of 6.6. In this paper we report on the application of the Fast Dose Calculator for photon dose calculations and its implementation on a GPU based architecture.

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II. Method and Materials

1. Monte Carlo

Based on the GEANT4 toolkit,[15,16] we developed a standalone Monte Carlo program to transport photons through a voxelized patient anatomy. The user-developed code provides descriptions for particles, physics processes, and materials to be simulated. In this work, the physics model configuration was taken from the protonEMLowEnergy case for the hadrontherapy example distributed with GEANT4 code version 4.8.3. Within the various packages for electromagnetic interactions, a model that takes into account atomic and shell effects and is applicable down to 250 eV was utilized. It makes ample use of parameterizations through evaluated libraries-EPDF97,[17] EEDL,[18] and EADL[19]- for final states for photon and electron collisions.

The GEANT4-based Monte Carlo code played two major roles in this study, as in our prior work.[8–10] First, the database of particle trajectories in water to be used by FDC was generated. And second, it was utilized to generate reference dose distributions in the voxelized anatomical phantom to evaluate FDC's performance. The results were normalized to the number of source photons.

2. Fast Dose Calculator for Photons

The basic features of the FDC algorithm for protons were described previously.[8,9] In this section, we describe the specific modifications needed for photon dose calculations.. The FDC method requires a database containing the trajectories of a few million photons traversing a water phantom. Using GEANT4, we simulated 10 million 2 MV photons impinging on a 300 × 300 × 600 mm^3 homogeneous water phantom. GEANT4 transported the photons by approximating the trajectories with discrete steps; the trajectories of all primary and secondary particles were recorded, including each step's path length, angle relative to the previous step, energy loss, and energy deposited. In our simulations, electron trajectories were terminated at a cutoff energy of 0.5 MeV and their residual kinetic energy was deposited locally. The deposited dose distribution in a heterogeneous media is then calculated by re-tracing the particles trajectories in the new media and scaling the length of particles, according to the material they are traversing. The scaling parameters for photons are given by the expression: $\alpha_M(E) = \lambda_M(E)/\lambda_{H_2O}(E)$, where α_M is the scaling parameter, λ is the photon mean free path, and the subscripts M and H_2O refer to material M and water, respectively. The photon mean free path was calculated as a function of energy, utilizing the cross section provided by GEANT4 for the Compton, pair-production and photoelectric processes. The same scaling parameters were used for electrons.

3. Patient Simulation and Field Definition

The irradiated volume was represented as a voxelized phantom based on the CT images of the thoracic region of a typical patient who had previously been treated for cancer at The University of Texas M. D. Anderson Cancer Center. The patient anatomy was represented by a computational phantom comprising 1,111,936 voxels, each having dimensions of 1.9 × 1.9 × 2.5 mm^3. Each voxel was assigned a material composition and density that corresponded to the Hounsfield unit value in the CT scan for that voxel, as described by Taddei *et al*.[20] In the present study we have limited our simulations to a beam of photons with a fixed energy. In the case of protons, the trajectory in water of a proton of energy E_i can be used to obtain the dose of a proton with an energy, $E < E_i$ in an inhomogeneous material. The proton loses energy mainly through a quasi continuous process of interactions with electrons in the media. Such a process leads to a continuous loss of energy, which allows to start using the track in water at the step where the energy corresponds to the energy of the proton that needs to be simulated in the inhomogeneous material. In the case of photons, they do not lose energy through a continuous process. Therefore to simulate a photon through an inhomogeneous material, we utilize a history in water of a photon with the same energy as the one that needs to be simulated. The simulation of photons with arbitrary energies requires an organization of the database of histories in water, which is not presented in this work. We used a circular beam of photons with 2 MV energy and a 2.5 cm radius. Similar results were obtained with different energies.

4. CUDA Implementation

A GPU-version of FDC, GFDC, was developed using the CUDA software platform[21] on a general purpose GPU graphics card (GEFORCE GTX 295, NVIDIA, Santa Clara, CA) with 1.79 GB of global memory as the hardware platform.[21] That GPU card used in the study holds 2 GPU units, with each unit holding 240 GPU cores. GFDC used only 10 million precalculated photon histories to minimize the time spend by the program reading the database. The GPU programming uses multiple computational threads. The total number of threads is controlled by the program. Threads are divided into blocks, so that the total number of threads is the number of blocks (N_B) multiplied by the number of threads per block (N_T). Each thread is treated as an independent computational unit, since the FDC is highly parallel,. Each unit re-traces one of the photon histories from the database of pre-calculated histories.

5. Computer Equipment

The processing times for the CPU version of FDC were recorded from calculations performed on a parallel computing cluster comprising 1,072 central processing unites (CPUs). The CPUs were contained in 134 nodes, each node comprising eight 64-bit CPUs (Xeon E5440; Intel, Santa Clara, CA) and operated at 2.83 GHz clock speed. Each node was provisioned with 16 GB RAM per node, which was shared among all CPUs on the node.

III. Results

Figure 1 shows the absorbed dose distribution versus depth in the patient anatomy in voxels along the beam central axis, y, plotted at z = 0 and x = 0 and 1.5 cm lateral to the beam central axis (i.e., z = 0 and x = 1.5 mm) as predicted by GEANT4, FDC, and GFDC. In addition, the difference between FDC/GFDC and GEANT4 doses divided by the maximum GEANT4 dose are plotted along the same axis.

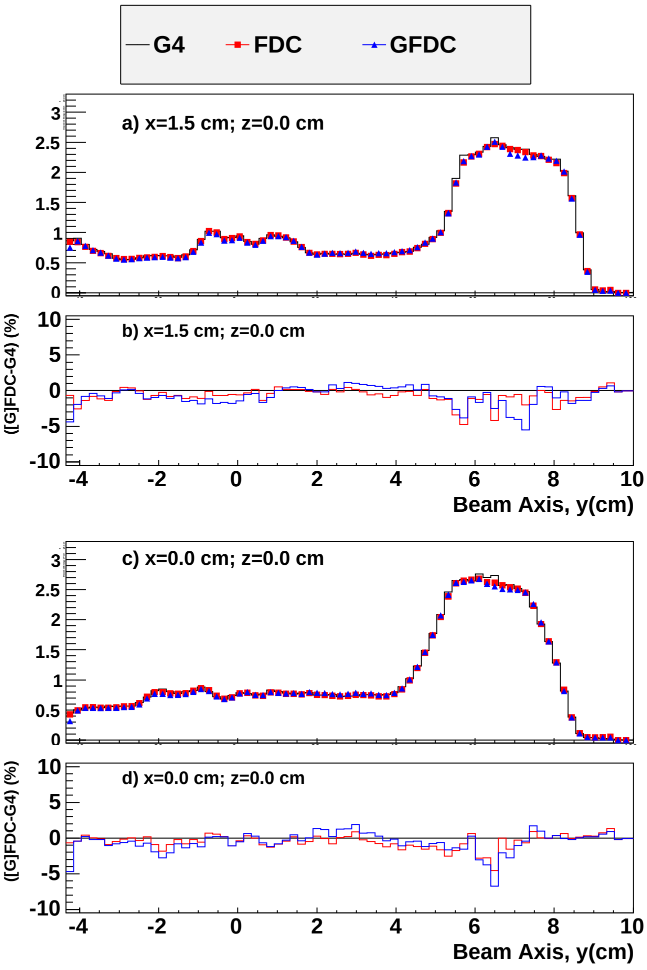

Fig. 1 Dose distributions along the beam axis, y, for (a) x = 1.5 cm and (b) x = 0 cm for z = 0 cm. Distributions were calculated with GEANT4 (G4: black line), FDC (red squares) and GFDC (blue triangles). The GEANT4-(G)FDC dose difference divided by the maximum GEANT4 dose is shown in panels (b) and (d) for x = 1.5 cm and x = 0 cm, respectively.

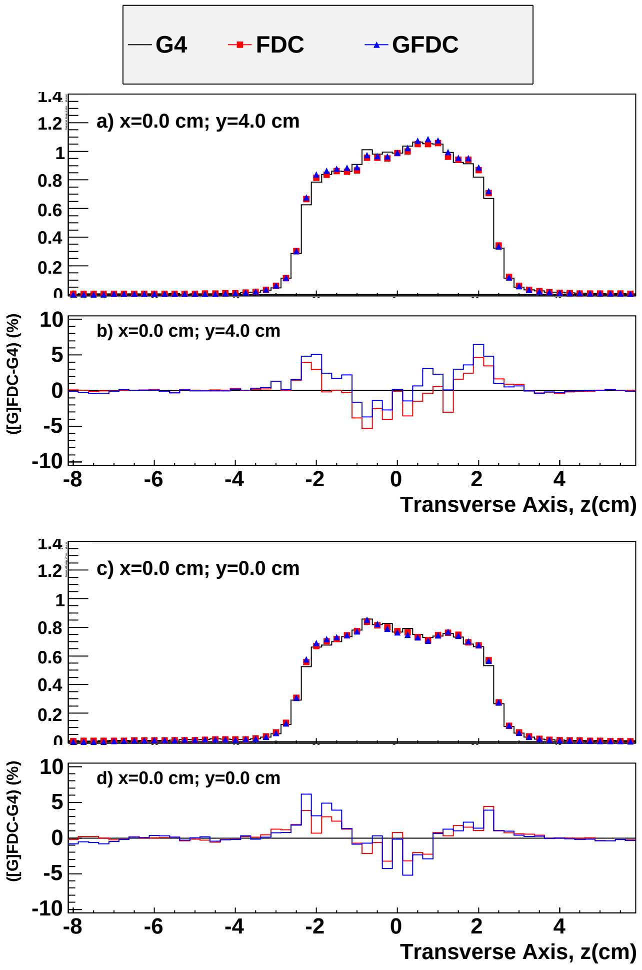

Fig. 2 Dose distributions along the z-axis, perpendicular to the beam, for (a) y = 0 cm and (c) y = 4 cm for x = 0 cm. Distributions were calculated with GEANT4 (G4: black line), FDC (red squares) and GFDC (blue triangles). The GEANT4-(G)FDC dose difference divided by the maximum GEANT4 dose is shown in panels (b) and (d) for y = 4 cm and x = 0 cm, respectively.

The equivalent distributions for an axis (z) perpendicular to the beam are depicted in **Fig. 2**. Good agreement was observed between the doses calculated by each code for both curves. The differences between FDC and GFDC can be mainly attributed to statistical fluctuations, since the databases of pre-calculated histories used by both implementations of the track-repeating algorithm were not identical. When GFDC and FDC were run on smaller data samples with identical sets of pre-calculated histories, the results were very similar, even though not completely identical. In that case, we attributed the difference to rounding errors on the different procesors. The calculation time per photon history was found to depend on the number of blocks (N_B) and the number of threads per block (N_T). The fastest calculation times were obtained for N_B=500 and N_T=320, for which 715,909 photon histories were processed per second utilizing the two GPU units on the graphics card. The CPU-based FDC on one CPU processed 2445 photon histories per second. Therefore the implementation of the FDC photon algorithm on a GPU card alone achieved a speedup of a factor of 103 with respect to the CPU-based implementation.

IV. Conclusions

A track-repeating algorithm for the simulation of the passage of photons through inhomogeneous media has been implemented on a GPU-based architecture for photons of fixed energy. The dosimetric accuracy of the algorithm was validated by comparing the results with those generated with GEANT4 Monte Carlo and CPU-based FDC simulations for a target representing the thoracic region of a patient being treated for cancer. The implementation of the photon track-repeating algorithm on a GPU architecture may allow for real-time dose calculations for photon radiotherapy with a system equipped with multiple GPU cards.

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