

## **SIMULATIONS OF AN IMAGE-GUIDED RADIOTHERAPY SYSTEM**

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### **ABSTRACT**

An *image guided radiotherapy system* consists of three major components: a radiation delivery unit, an irradiated object (patient) and an imaging unit. While simulations of each of the components can provide some information, simulations of an integrated system provide another level of information. These simulations enable, for example, optimization of the system, improvement of the imaging feedback, and simulation of the dose delivery allowing later comparison to the real delivery to detect errors.

A Monte Carlo model was created with the MCNP Monte Carlo code to enable simulation of an integrated image guided radiotherapy system with the helical tomotherapy system as an example. All three basic components of the system were modeled with the highest possible accuracy. Even though most of the geometric information was available, some details about the detector geometry were not known and significant effort had to be put into “reverse engineering” the system.

Data from the Monte Carlo model were compared with experimental data from the phantom and detector. Excellent agreement was observed with the measurements in the phantom, which enabled tuning of the parameters of the incident electron beam. On the other hand, agreement between the measured detector signal and calculated dose deposited in the detector was in lesser agreement. Several important characteristics have been determined from the system, like the photon and electron spectra at the exit of the source and in the detector, the amount of electron contamination, and the amount of the primary/scattered signal.

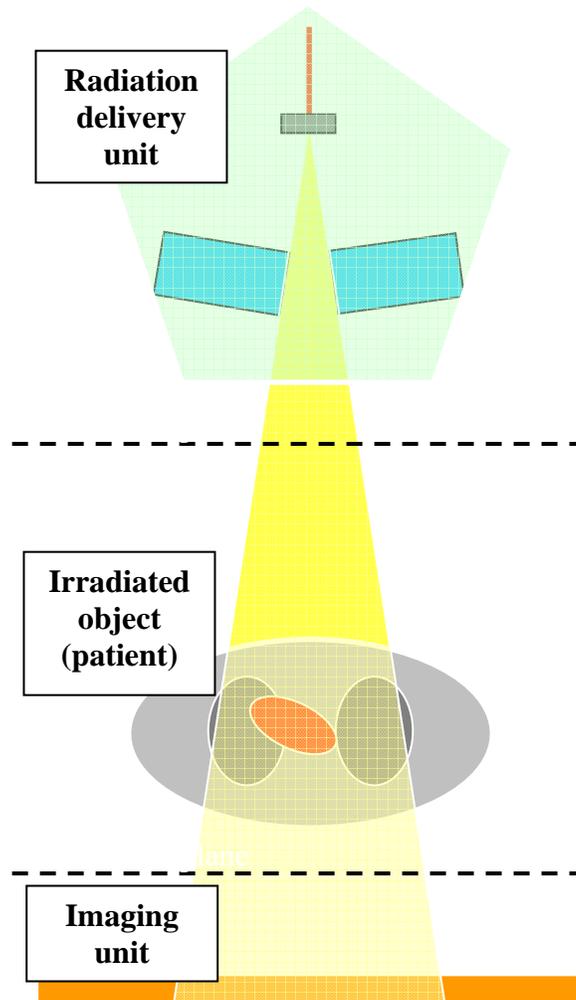
Monte Carlo simulations of an image guided radiotherapy system are very useful for determining characteristics of the system that can not be measured and are helpful for improving the system functionality. However, extensive effort has to be put into commissioning of the system, especially since some of the design details might not be available.

*Key Words:* Monte Carlo simulations, image guided radiotherapy system,

### **1 INTRODUCTION**

An *image guided radiotherapy (IGRT) system* consists of three major components: a radiation source unit, an irradiated object (patient) and an imaging unit. While simulations of each of the components can provide some information, simulations of an integrated system provide another level of information, which enables, for example: optimization of the system,

simulation of the delivery and later comparison to the real delivery to detect errors, improvement of the imaging feedback etc.



**Figure 1:** Components of an image-guided radiotherapy unit. Monte Carlo simulations require accurate and detailed modeling of all three principal components – the radiation delivery unit, the irradiated object (patient) and the imaging unit.

Accurate Monte Carlo transport modeling is very important to characterize and understand the physical and dosimetric characteristics of clinical radiation therapy units. Therefore characterization is the main area where Monte Carlo simulations have been used so far for both photon and electron clinical beams (Petti, Goodman et al. 1983; Mohan, Chui et al. 1985; Chaney, Cullip et al. 1994; Ding and Rogers 1996; Ebert, Hoban et al. 1996; Lee 1997; Ma, Mok et al. 1999; van der Zee and Welleweerd 1999). Substantial effort has been put into construction of beam models, which can be used instead of phase space files (Ma, Faddegon et al. 1997; Schach von Wittenau, Cox et al. 1999; Chetty, DeMarco et al. 2000; Deng, Jiang et al. 2000). Monte Carlo models/codes have also been extensively validated through comparison to measurements (Faddegon, Ross et al. 1991; Kapur, Ma et al. 1998; Verhaegen, Das et al. 1998; De Vlamynck, Palmans et al. 1999; Ma, Mok et al. 1999; Li, Pawlicki et al. 2000; Sheikh-Bagheri, Rogers et al. 2000; Sheikh-Bagheri and Rogers 2002) and different codes has been inter-compared (Jeraj, Keall et al. 1999; Siebers, Keall et al. 1999; Sempau, Sanchez-Reyes et al.

2001). Furthermore, Monte Carlo calculations have also been used for validation of non-Monte Carlo treatment planning systems (Jeraj, Keall et al. 1999; Siebers, Keall et al. 1999; Sempau, Sanchez-Reyes et al. 2001). Most of Monte Carlo simulations of clinical radiotherapy units have been based or performed with the BEAM/EGS4 code (Nelson, Hirayama et al. 1985; Kawrakow 2000; Kawrakow and Rogers 2000), but other codes like MCNP (Briesmeister 2000), PENELOPE (Sempau, Sanchez-Reyes et al. 2001) have also been used. Monte Carlo methods are also very valuable for commissioning of a treatment unit since they enable various sensitivity studies, which can test the effects of changing different parameters on output characteristics of a treatment unit. Commissioning of treatment planning systems, either being based on non-Monte Carlo dose calculation (e.g. convolution/superposition) or Monte Carlo dose calculation is another important area where Monte Carlo calculations can be of great help. Furthermore, Monte Carlo simulations are very important in cases where measurements are difficult, inaccurate or even impossible. In order to achieve the highest possible accuracy of the calculations and therefore reliable answers, a Monte Carlo model should be as accurate as possible, preferably based on engineering drawings.

The main purpose of this paper is to review diverse use of Monte Carlo simulations for modeling of an image-guided radiotherapy system. Various examples of characterization, optimization and commissioning of the treatment unit including the treatment planning system are described.

## 2 MATERIALS AND METHODS

As an example of an image-guided radiotherapy system - helical tomotherapy treatment system (Mackie, Holmes et al. 1993) - is presented here. Helical tomotherapy (Figure 2) is the first treatment unit dedicated to intensity modulated radiotherapy (IMRT) and a fully integrated image guided radiotherapy (IGRT) system. It was developed and constructed at the University of Wisconsin – Madison and TomoTherapy Inc., Madison, WI.



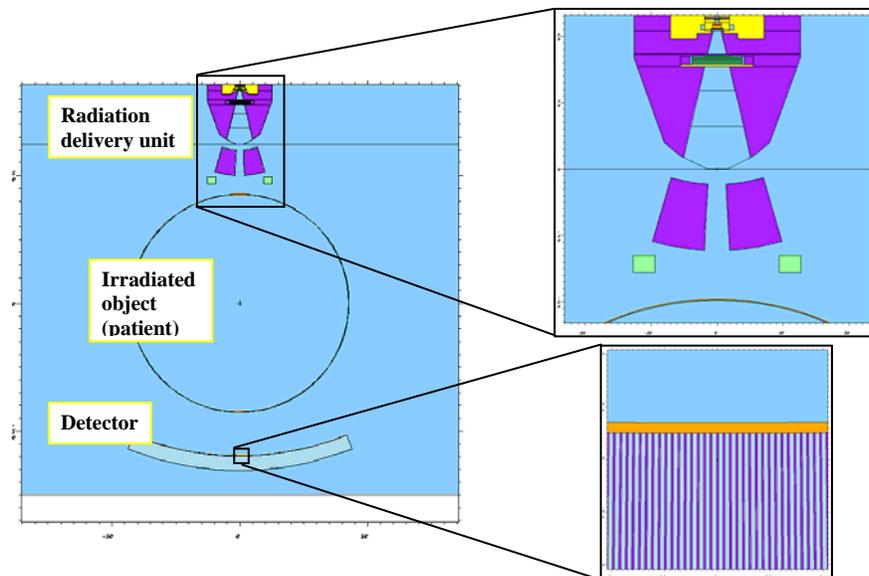
**Figure 2:** The helical tomotherapy system used as a model of an image-guided radiotherapy system in this work.

The basic radiation source of the helical tomotherapy system is a linear accelerator mounted on a rotating gantry similarly to a CT scanner. The radiation is delivered to a patient in a helical way, obtained by concurrent gantry rotation and couch/patient travel. The treatment field is modulated with a 64-leaf binary multileaf collimator. The Monte Carlo model of the *radiation source unit* and *imaging unit* was based on the original engineering drawings with original material specifications. The model was not compromised in any way for accuracy. The *irradiated object* model was also based on realistic data. All the phantoms used in this work were water phantoms simulated as boxes of  $30 \times 80 \times 50 \text{ cm}^3$  with typical voxel dimensions of  $0.5 \times 0.5 \times 0.5 \text{ cm}^3$ . The patient models were created automatically from computed tomography (CT) data. The Monte Carlo model of the helical tomotherapy system with some details on the radiation delivery and imaging units is shown in Figure 3.

The Monte Carlo code used in our studies was MCNP4c3 with its default cross section libraries. In the simulations, the only free parameters that had to be tuned were characteristics of the incident electron beam on the target (spatial, angular and energy distribution). Several functional dependencies were tested and their parameters tuned in order to get the best agreement with the experimental dosimetric data (depth dose curves, profiles, output factors). The incident electron beam on the target was modeled to be Gaussian in energy and spatial distribution with constant, but limited angular distribution.

Three phase space scoring planes were used in the complete Monte Carlo model of the helical tomotherapy unit. The first one was placed between the treatment head and multileaf collimator, the second one after the multileaf collimator (before the patient) and the third one after the patient (before the detector). The following variance reduction methods were used in order to increase the simulation speed: cell-dependent energy cut off, geometry importance splitting/Russian roulette, and bremsstrahlung splitting.

The Monte Carlo model was first commissioned using the measured dose distributions using a standard commissioning procedure. During this procedure the incident electron beam was characterized. In all subsequent studies these parameters were used.

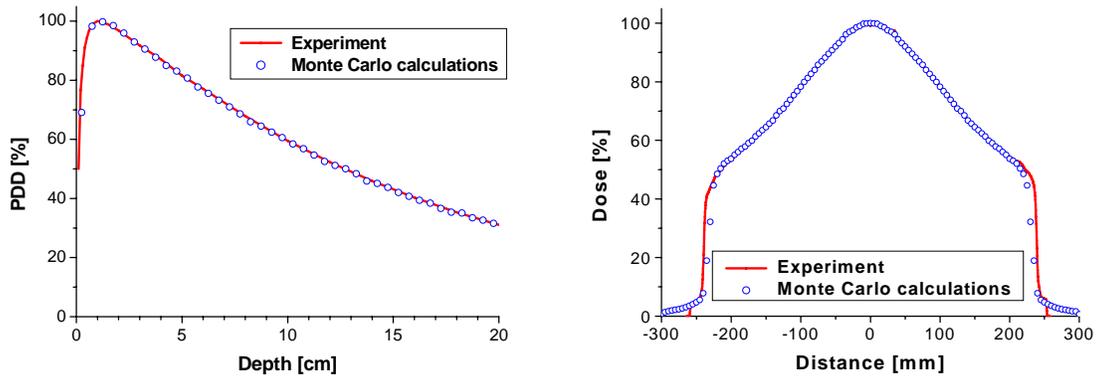


**Figure 3:** The complete helical tomotherapy Monte Carlo model (a) and two blow-up details of the radiation delivery unit (b) and detector (c).

### 3 RESULTS

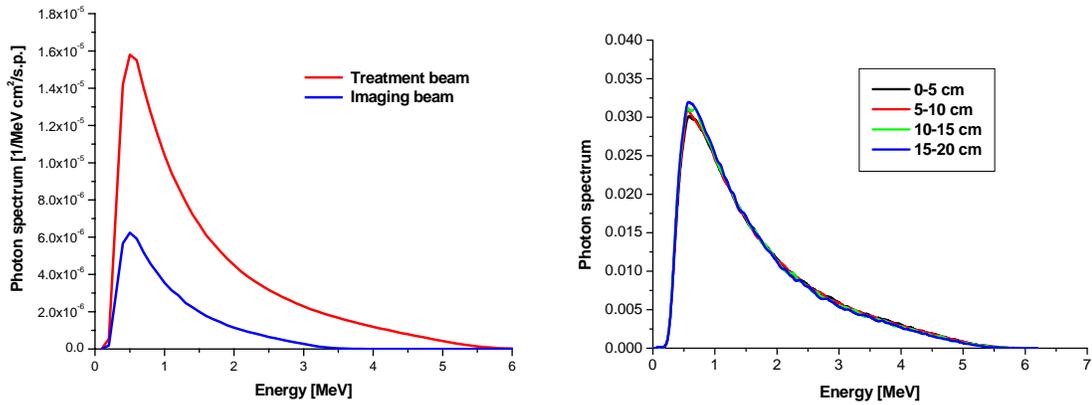
#### 3.1. Radiation source unit

The prerequisite for treatment simulation of an IGRT system is accurate model of the radiation source and delivery unit. Before using the model, the model has to be first commissioned and then tested on the experimental data. Examples of the model prediction and experimental results are shown in Figure 4 for depth dose curves and lateral dose profiles. The best match with the experimental data was obtained for the following source parameters: incident electron energy of 5.7 MeV with the Gaussian energy spread. The spread was found to be rather unimportant, and was kept at approximately 5% of the nominal energy. The radial dependence of the incident electron beam was also modeled to be Gaussian with the FWHM of 1.4 mm (Jeraj, Mackie et al. 2004). It should be emphasized that the lateral profiles for the helical tomotherapy system differ significantly from other conventional treatment systems and exhibit characteristic conical profile, typical of the unfiltered bremsstrahlung intensity profile. The reason for observance of this profile is absence of the flattening filter, which is in the case of an IMRT-dedicated system redundant. Furthermore, inclusion of the filter in any IMRT system unnecessarily reduces the available intensity for approximately a factor of 2 for lower energies (~ 6MV) and even more for higher energies because of the increased bremsstrahlung differential cross-section in the forward direction.



**Figure 4:** Comparison of the calculations and measurements for central axis depth dose curves and profiles. Incident electron distribution on the target was tuned until the satisfactory agreement between the simulations and measurements was achieved.

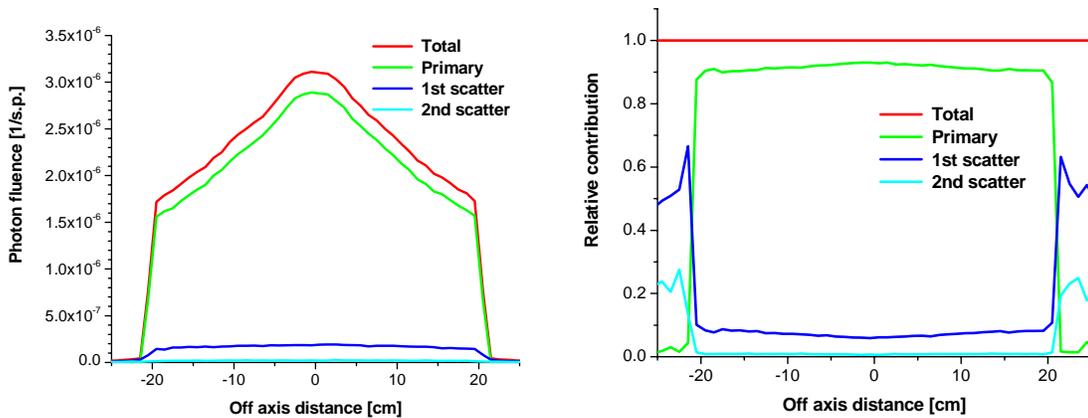
One of the important characteristics and main results of the radiation source unit modeling is the photon spectrum. In the helical tomotherapy, as in other image-guided radiotherapy system, both the treatment and imaging beam have to be treated separately. In case of helical tomotherapy, the imaging beam is produced by lowering the energy of the accelerated electrons thus producing a softer spectrum (Figure 5). The typical treatment and imaging energies are approximately 6 and 3.5 MV, respectively. The helical tomotherapy has another specific spectrum characteristics – the central-axis beam hardening (or equivalently off-axis beam softening), typical for conventional treatment systems (typically between 30 and 40%) is not present because of the afore-mentioned absence of the flattening filter (Figure 5).

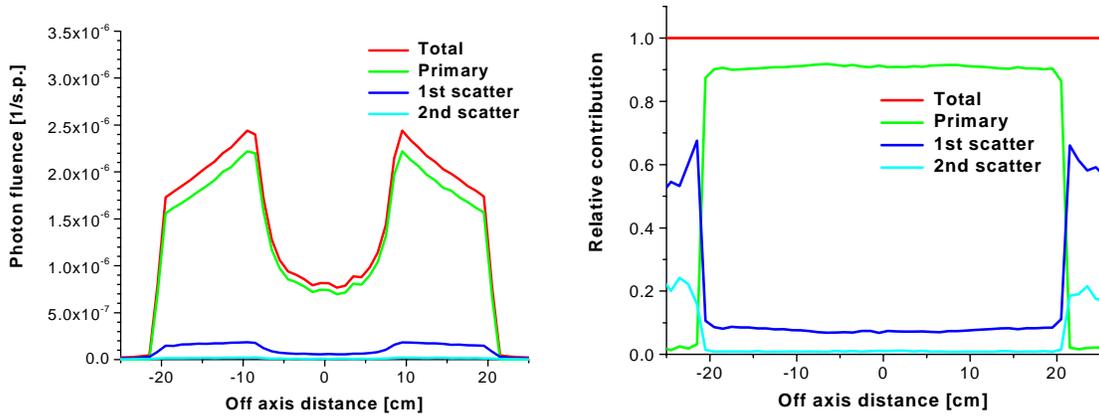


**Figure 5:** Photon spectra for the treatment and imaging beams (left). Normalized spectra as a function of the distance from the central axis (right). Spectrum hardening on the central axis, compared to the edge of the field, typical of other radiotherapy systems, is not present here because of the absence of a flattening filter.

### 3.2. Irradiated object (patient)

The main advantage of the IGRT system over the conventional radiation treatment systems is imaging feedback enabled by the on-board detector unit. In order to test the IGRT simulation loop an object was irradiated and its effect on the detector signal was examined. Photon fluence and relative contributions of primary and scatter radiation are shown in Figure 6 for the case with and for the case without the object in the beam. It is interesting to note that the amount of scatter is relatively low because of the narrow fan beam geometry used for imaging. Furthermore, the amount of scatter does not increase significantly with the patient in the beam and remains rather constant across the beam.

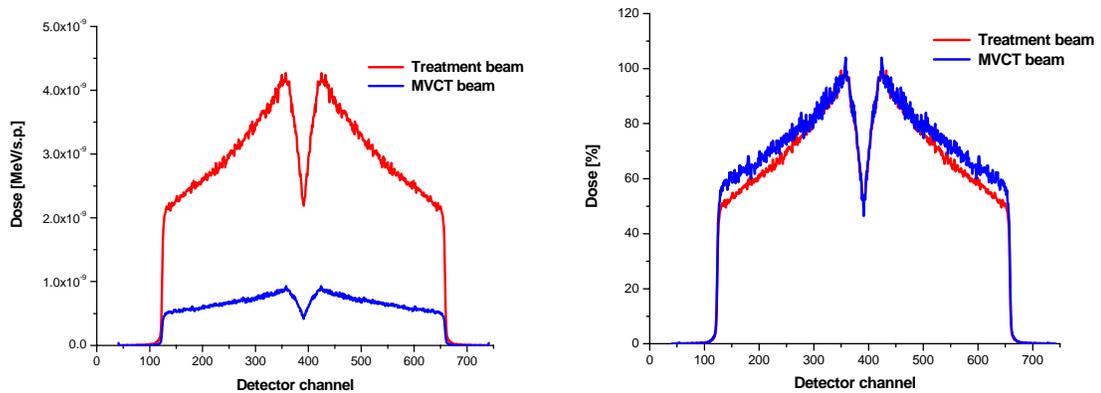




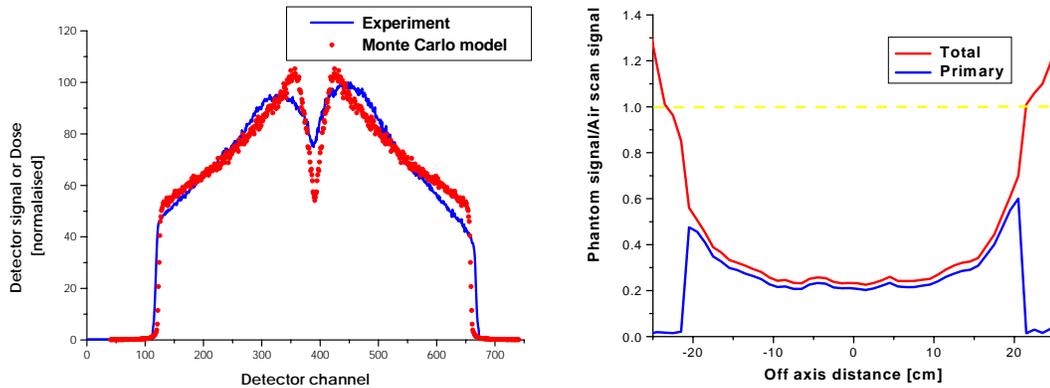
**Figure 6:** Photon fluence and relative contributions of primary and scatter radiation for a head and neck treatment case. Large fraction of primary radiation indicates that photon beam model based on primary component might be very effective.

### 3.3. Radiation detector unit

Detector systems used for imaging can also serve for treatment delivery monitoring. It should be mentioned that exact simulation of the imaging chain is not possible in the model, since only the radiation transport part is considered, while the detection efficiency, electronics readout and noise associated with electronics have to be brought in by other means. A comparison between the detector signals for a treatment and imaging beam are shown in Figure 7.



**Figure 7:** Comparison between the detector read-out and calculated absorbed dose in each of the 700 detector channels (left). Treatment verification with included additional information about the primary detector signal, which can help in improving image reconstruction (right).



**Figure 8:** Comparison between the detector read-out and calculated absorbed dose in each of the 700 detector channels (left). Treatment verification with included additional information about the primary detector signal, which can help in improving image reconstruction.

Note significantly lower signal (per incident photon) for the imaging beam compared to the treatment beam. In addition, the signal off-axis drop is smaller for the imaging beam compared to the treatment beam because of the higher energy. A comparison between the detector signal as read out and the absorbed dose in each detector channel, together with photon and electron fluence is shown in Figure 8. The significant decrease in the center of the beam is due to the detector structure (alternating W plates and air cavities), which causes the in-focused part of the detector to have much lower photon beam conversion efficiency and thus leading to a lower signal. While the overall agreement between the measurements and simulation is relatively good and all important features of the signal profile are visible, further tuning/commissioning is needed. The main reason for the discrepancies is probably a non-perfect alignment of the detector in the real system. In addition, the relationship between the detector signal (charge) and the absorbed dose (or any other quantity) would need to be determined, to best represent the real situation.

## 4 CONCLUSIONS

Monte Carlo simulations of an image guided radiotherapy system are very useful for determining characteristics of the system that can not be measured, and are helpful for improving the system functionality. However, extensive effort has to be put into commissioning of the system, especially since some of the design details might not be available.

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