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Realistic Monte Carlo simulation of the INOR Therapy Unit and electron contamination calculation for a board ^{60}Co Beam

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The therapy beam from the ^{60}Co unit used at INOR was modeled using the EGS Monte Carlo radiation transport system. The rectangular geometry of the collimating system is fully simulated and the effect of considering the outer collimator a set of stepped collimators rather than a collimator with a smooth surface was estimated. Variance reduction techniques are widely used to increase the efficiency of our calculations. Even for the dose at very shallow depths (<0.5 mm) the agreement of our calculations with the experimental results is very good. The shift of peak dose to about 1 mm due to electron contamination is observed.

1. Introduction

For treatment planning in ^{60}Co cancer radiotherapy it is desirable to have a detailed knowledge of the surface dose, since one of the major advantages of ^{60}Co beams compared to lower energy X-Ray beams is the skin sparing effect. For a pure ^{60}Co beam the dose at the surface is low and builds up to a peak at about 5 mm depth. In practice the buildup curve is not as good as expected. Electron contamination of broad ^{60}Co therapy beams can increase the maximum dose by up to 15% and can shift the depth of dose maximum from 5 mm to 1 mm.There have been several theoretical and experimental investigations of this effect (e.g., [1] to [4] and references therein) and it is now well established that electrons are the major contaminants.

The aim of this work was to develop and benchmark a sophisticated Monte Carlo simulation model of a ^{60}Co therapy unit to obtain a realistic beam of the therapy head that allows us to calculate the contribution of electrons to the surface dose. From a physical point of view, it was impossible before to achieve a good description of absorbed dose near the surface. Moreover, there is also interest to see the clinical impact of our results on radiotherapy planning and treatment of patients in close geometry, where electron contamination makes a big contribution to the dose.

Starting point of our model was the work of D.W.O. Rogers et al.[6], who calculated the electron contamination of a ^{60}Co photon beam. In this extensive work, the collimating system of the therapy unit was modeled assuming cylindrical geometry rather than rectangular as it is in fact. The outer collimator was solid lead and depth-dose-curves were calculated by

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scoring the planar fluence at the patient surface and folding these with pre-computed depth-dose-curves for normally incident monoenergetic particles. They showed that there are many sources of electron contamination (collimators, air, source capsule) which vary in importance as a function of source-to-surface distance(SSD).

They have also studied the effect of filters and magnets on electron contamination. Their calculations are shown to be in good agreement with experimental results except very close to the surface (< 0.5 mm) where the calculations underestimate the measurements.

In this work we are interested on an exact simulation of the INOR radiotherapy unit (THERATRON 780-C) considering that the geometry of the collimating system is rectangular rather than cylindrical and that the outer collimator is a set of stepped collimators separated by air gaps. With this model we calculate then absorbed dose, evaluating electron contamination for a close irradiation geometry.

2. Calculation

In our model we have used the Electron-Gamma-Shower Monte Carlo system[5] (EGS) to simulate the transport of electrons and photons. For photons EGS takes into account photo-electric, Compton and pair production interactions and accounts for the slowing of electrons, the creation of secondary photons and electrons, and the multiple scattering of electrons.

The model assumed for the INOR radiotherapy unit is shown in Figure 1 Their main parts are: source and collimating system.

To take into account the contribution of electrons and scattered photons generated in the source, it was essential to model fully the ^{60}Co source capsule. In the code SOURCE a typical capsule has been modeled (see Figure 1). A real source capsule contains radioactive ^{60}Co pellets that are simulated as low density iron and the heavy metal sleeve around the capsule is modeled by an equivalent thickness in gcm^{-2} of Fe. The ^{60}Co radiates uniformly into 4π.

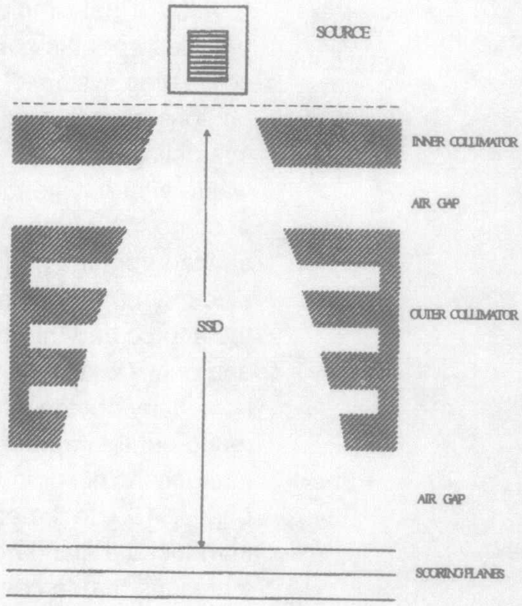

Figure 1. *Model Using for simulating the ^{60}Co unit. There is rectangular symmetry. All dimensions and materials are variable.*

The collimating system of the INOR radiotherapy unit is described below. Two collimators are used- inner fixed and outer moveable. In our model the inner collimator is solid tungsten, and fixed to obtain a beam of 35x35 cm2 area at an SSD of 80 cm. It is 1.50 cm from the capsule, 6.2 cm thick, with inner and outer openings of 1.1 cm and 2.4 cm. The outer collimator is lead, moveable and often made of a series of leaves. The nominal field size at a given SSD is defined by the straight line joining the center of the front face of the ^{60}Co capsule and the outer edge of the outer collimator.

The calculation was divided into four steps to increase efficiency. In the first step we simulate the emission from the ^{60}Co capsule. The phase space parameters of 2 million particles emerging from the right face of the capsule are stored in a binary format and used as input for the rest of the simulation.

In the second step we use the large set of source particles as input to the code COLIMAT that simulates the passage of the particles from the source to the end of the collimating system. The most part of the photons pass through the air without interacting, and a direct simulation would waste a lot of time following these primary photons. So we have that for 10^6 particles emitted from the source (only 0.5% are electrons) 93% photons interact between the source and the exit of the therapy head and only 7% did not. We have also counted how many photon interactions occured and found out that aproximately 99.98% interactions occur in the collimators and only 0.016% in the air. Therefore, we have introduced a standard variance reduction technique that forces all the primary photons from the source to interact in the air regions between the source and the end of the second collimator. The model takes into account this unphysical large number of interactions by reducing the weight of the resulting particles. It is also intersting to notice that from the incoming particles only 6.95% come out through the collimating system and that is close to the number of photons that do not interact at all. Again from these outcoming particles only 0.83% belong to initial photons that have interacted or to its offsprings For this part of the simulation we have used the default step size algorithm for the electron transport.

In the third step the code AIRGAP is used to simulate the passage of the particles, leaving the collimating system through the air to the patient surface. Also hier primary photons are forced to interact. To make efficient use of the large amount of time spent calculating the passage of particles through collimating system we reuse the same particles leaving the outer collimator for each SSD being considered. Electrons are followed until their energy drops under 10 keV (ECUT=0.521 MeV) and a reduced step-size ESTEPE of 4% is used.

In the last step we compute the dose deposited by the particles of the ^{60}Co beam in a tissue-equivalent medium simulating the full radiation transport in this medium. We have used a modified version of the XYZ code named DOSIS for doing the simulation at this step. The reduced electron step-size ESTEPE is 4%. Since we are interested on the dose in the builup region we employed the exponential transform of the photon path length. So, more photons interact in this region and the efficiency for dose calculation is improved there. Through several calculations we found necessary to choose the value -10 for the biasing constant C. The biasing is re-stricted to photons that are directed along the z-axis (cosz > 0). Due to this severe biasing, backscattered photons with very large weights can reach the region of interest and increase the variance. For this reason we have used a *splitting* technique for photons if they threaten to enter the region of interest. PCUT = 0.001 MeV and ECUT = 0.521 MeV so that electrons are tracked down to 10 keV and photons to 1keV in medium.

The complete beam used for the dose calculation is built by considering all the possible ways photons can interact in the whole system. Having this in mind we took 10^6 particles emerging from the source and simulated the passage of the beam through the collimating system without forcing photon interactions and then the other 10^6 particles from the source letting photons interact in the air regions between the source and the outer collimator. In this way we obtain two outputs, which are then used as input through the air gap between the therapy head and the patient plane. At this stage we repeat the same procedure for both beams (forced and no forced) simulating the passage through the air gap once forcing and then no forcing. As result we obtain four different situations†: no forcing in both the collimating system and the air gap, forcing in the collimating system and no forcing in the air gap, the reverse situation of the latter and forcing in both regions. With this algorithm we are interested in collecting information of the whole system that could be useful to obtain a realistic composition of the ^{60}Co beam.

3. Results

3.1. Source output

We have simulated the photon emission of the source capsule and the interaction of the photons within it. The scattered photons and the electrons contribute 64% and 0.43% respectively of the particles leaving the front face of the source. The electrons have an average energy of 580 keV.

Figure 2 shows the angular distribution of the photons leaving the source. As can be seen, the forward direction is preferred. The iron encapsulation around the ^{60}Co nucleus allows the source to emit photons mostly forwards.

3.2 Collimators

Two million particles from the source were used for simulating the passage of a ^{60}Co source beam through the collimating system and the outcoming particles represents only 7% of the input.

Neglecting scattered photons from the collimators we found that their contribution amounts up to 28% of the photon fluence. Including collimator scattering this component goes up to about 35% of the photon fluence (Figure 3).

To avoid overestimating the dose in the buildup region it was found necessary to simulate the outer collimator as a set of stepped collimators. Regarding the outer collimator as solid and with smooth surfaces produces to much scattered photons and secondary electrons. A comparison between both cases is shown in Figure 4.

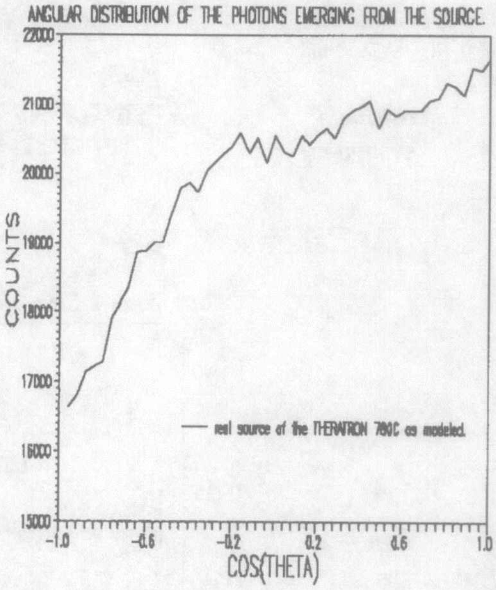

Figure 2. *Angular distribution of the emitted photons from the source. Forward direction is preferred.*

The most time of the calculation at this step is spent simulating the transport in the collimators. Ignoring this reduces the computing time in a PENTIUM 90 from 1h17min to 5 min.! (These values depend strongly on the chosen set of transport parameters ECUT and PCUT).

3.3 Passage through the airgap

The airgap between the head and the patient is both a source and an attenuator of electrons. We have done the simulation for two values of ECUT, 611 keV and 521 keV.

The full depth dose curve for both cases remains the same except for the dose at the surface, where an increase of the surface dose for ECUT=521keV is achieved.

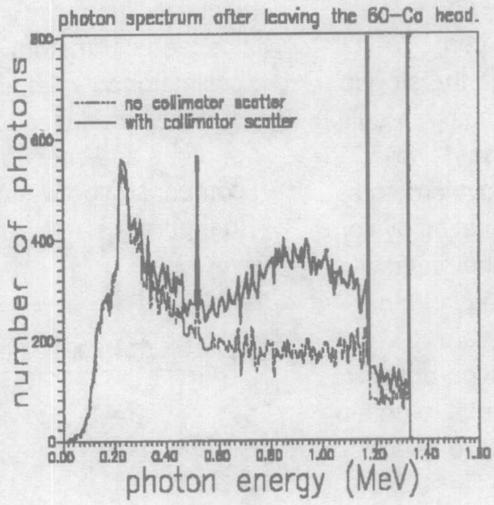

Figure 3. *Photon spectra, just after leaving the* ^{60}Co *unit with and without the effect of collimator sacttering*

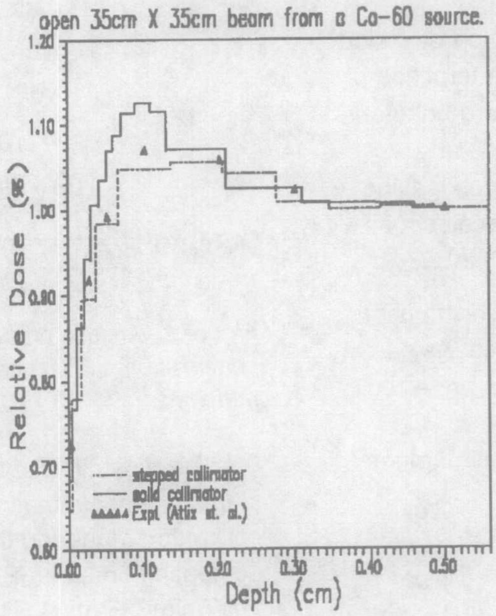

Figure 4. *Histograms show the effect of taking the outer collimator once as a solid collimator and as a set of stepped collimators. Measurements of attix et. al. are used*

3.4 Comparison with experimental results

Calculations of the complete depth-dose curves for the simulated ^{60}Co were compared with measurements reported by other authors (see Figures 5 and 6).

Figure 5 compares the calculation with the results of Attix *et al.*[3]. They measured absorbed dose in the buildup region for polystyrene until very shallow depths (15 μm). Our results reproduce the experiment with accuracy even for this shallow depth. The position of the peak dose lies by 1mm in agreement with the experiment.

Rogers *et al.*[6] have achieved in their calculations a very good agreement with Attix results except at the very surface (< 0.5 mm), where their calculations underestimate the dose.

Their calculated results for the relative dose at 15μm are roughly 30% low in all situations considered [6]. Our calculations show a discrepancy of roughly 7% at this depth.

In Figure 6 comparison with experimental results from INOR[7] and from Leung et al.[11] are shown.No data for 15 μm are reported.

As in figure 5 the agreement of our calculations is also good. In spite of discrepancies between both measurements due to differences in the experimental setup, the general behavior is similar to our calculations.

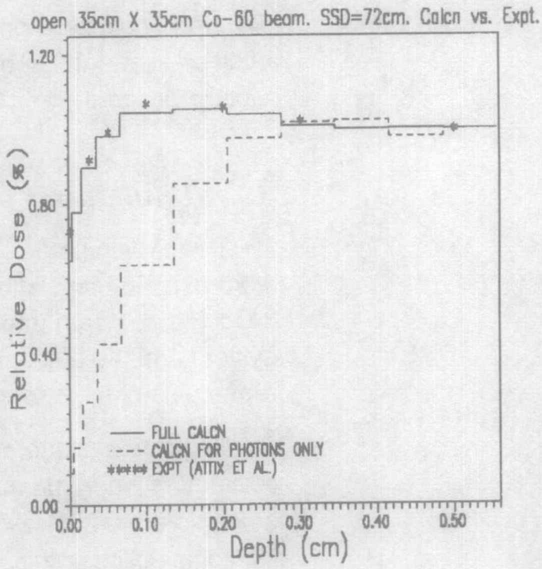

Figure 5. *Comparison of the calculated and measured*[3] *depht-dose-curves at an SSD of 72 cm. The calculated dose for photons only us also shown.*

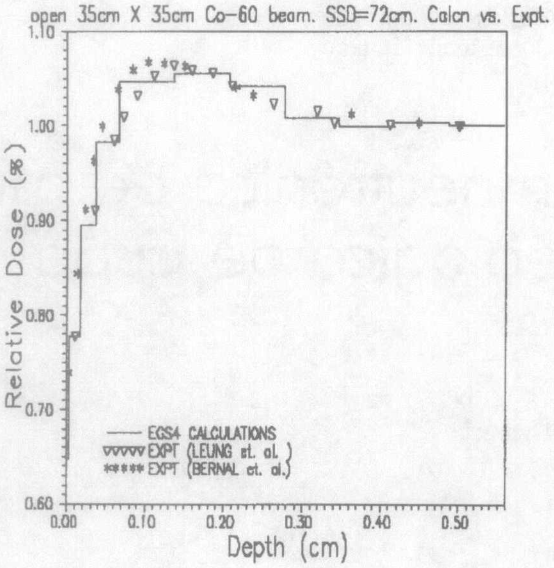

Figure 6. Comparison of our calculations with the experimental results of two other authors[7,11]

4. Conclusions

Calculations of the depth dose curve for a broad ^{60}Co therapy beam show that electron contamination is responsible for the increase of the surface dose and the shift of the peak dose near to the surface. It is necessary to simulate the second collimator as a set of stepped collimators to avoid overestimating the dose in the buildup region. Realistic simulation of the therapy unit and extensive use of variance reduction techniques allowed us to obtain very accurate results for the depth- dose-curves even for regions very close to surface with 4.7% statistical uncertainty at 1.6 mg cm^{-2} (0.0015 cm depth) and for depths over 0.05 cm less than 2%. Increasing the superficial bin two times reduces the statistical uncertainty to 1.72%.

5. References

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