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MONTE CARLO SIMULATIONS IN SKIN RADIOTHERAPY

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ABSTRACT

The primary goal of this work was to develop a procedure for calculation the appropriate filter shape for a brachytherapy applicator used for skin radiotherapy. In the applicator a radioactive source is positioned close to the skin. Without a filter, the resultant dose distribution would be highly nonuniform.High uniformity is usually required however. This can be achieved using an appropriately shaped filter, which flattens the dose profile. Because of the complexity of the transport and geometry, Monte Carlo simulations had to be used. An ^{192}Ir high dose rate photon source was used. All necessary transport parameters were simulated with the MCNP4B Monte Carlo code. A highly efficient iterative procedure was developed, which enabled calculation of the optimal filter shape in only few iterations. The initially non-uniform dose distributions became uniform within a percent when applying the filter calculated by this procedure.

1 INTRODUCTION

Skin cancer can be treated with different methods; the most commonly used rely on so-called superficial orthovoltage X-ray units (photon beam energies of several hundred MeV). An alternative is to use a radioactive source close to the target volume (see Figure 1). While this method has many advantages such as favourable depth dose characteristics, it is associated with non-uniform dose distribution at the surface. The non-uniform dose distribution, in the upper layers of the tissue, is the consequence of the short distance from the radiation source. However, it is usually desirable to have a uniform dose distribution. This can be achieved for example, by putting a flattening filter between the source and cancerous tissue [1]. To find an optimal filter shape Monte Carlo simulations of the experiment had to be performed. The MCNP4B [2] Monte Carlo transport program was used to calculate the required parameters. Lead was chosen as an appropriate material for the filter.

2 MONTE CARLO SIMULATIONS

2.1 Physical interactions

In our example a photon source (in a catheter) was defined at 15mm distance from the surface. This was enclosed by cylindrical lead shield and the dose distribution was calculated in a homogeneous water phantom. 2cm of the water phantom was sliced in 2mm thin layers divided radially from 1mm to 40mm. Between the phantom and the shield a 2mm thin perspex layer was inserted. The described geometry set-up is shown in Figure 1.

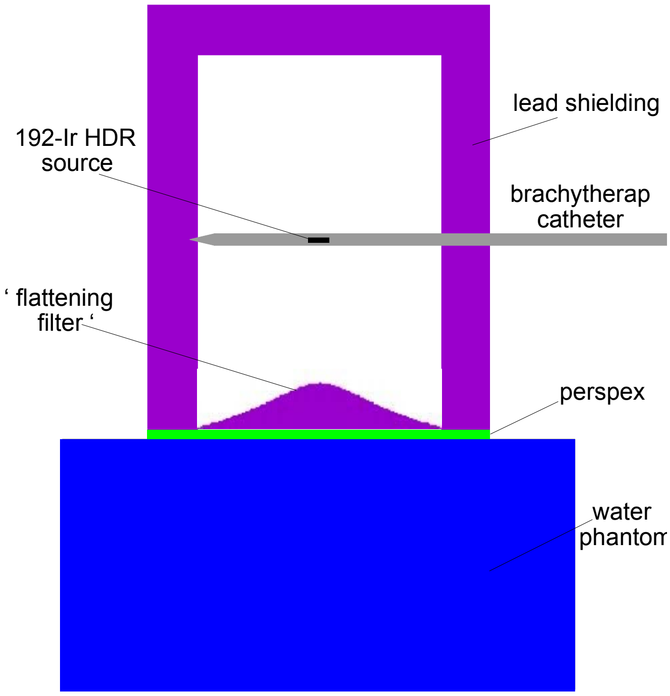


Figure 1 Geometry of the experiment. The shield and flattening filter are from same material. It is not shown how the water phantom is dissected.

In our work a high dose rate brachytherapy ^{192}Ir photon source was used [3]. An accurate energy spectrum was used in Monte Carlo calculations. Its distribution is shown in Figure 2. Because the source energy spectrum extends up to 620 keV including numerous lines, theoretical approach is not practical. From Figure 2 it can be seen that the main peak is at 320keV. In the photon transport through the different materials (lead, perspex, water) only the photoelectric effect, Rayleigh and Compton scattering were simulated because of the low energy of the source [4,5,6,7], although other interactions (e.g. pair production) were allowed in Monte Carlo simulation as well. In the electron transport MCNP considers the electron scattering and all secondary particles, born in different interactions.

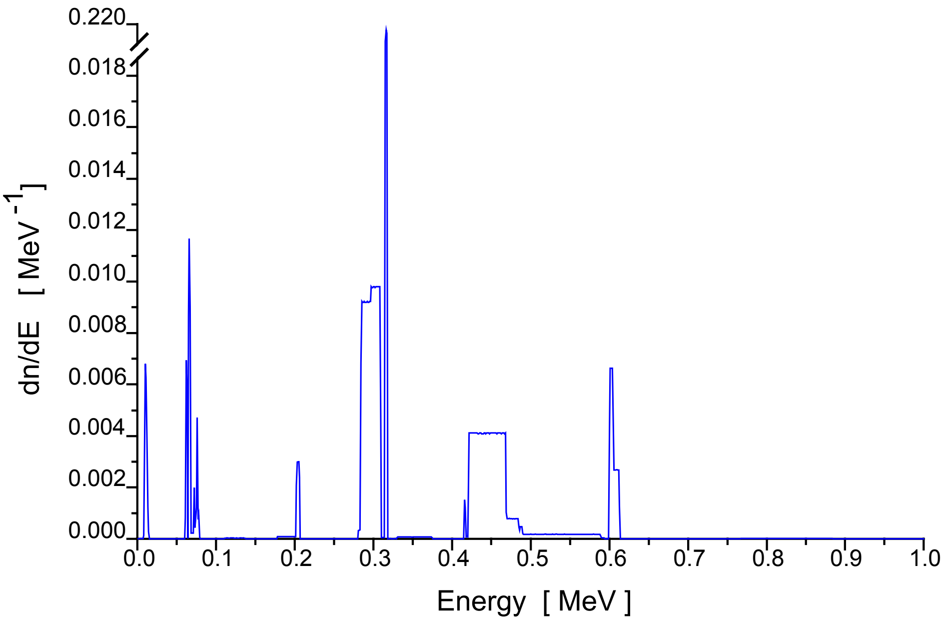


Figure 2 **The energy distribution of the ^{192}Ir source.**

An example of the type of interactions that occur in our simulation is presented in Table 1. From Table 1 it can be seen the importance of the interactions, which gives a better physical insight of the particle transport in the simulated geometry.

Table 1 **Relative number of the physical processes in different materials for photons and electrons (in percentage)**

MATERIAL \ PHOTONS	CAPTURE	BREMSTRAHLUNG	FLUORESCENCE	
Lead filter	63.49	3.41	33.10	
Water phantom	83.11	16.89		
Perspex layer	80.00	20.00		
MATERIAL \ ELECTRONS	PHOTON AUGER	COMPTON RECOIL	PHOTO ELECTRON	KNOCK-ON
Lead filter	1.76	10.00	84.05	4.19
Water phantom		97.34	0.92	1.74
Perspex layer		95.44	0.40	4.16

2.2 Theoretical examples

Initially our calculations were done with the assumption of the constant attenuation coefficient. However, to be more accurate, the dependence of the attenuation coefficient with depth had to be accounted for. A monodirectional photon source with a realistic ^{192}Ir energy spectrum was used and the flux through different thin lead layers was calculated. The thickness of the layers varied from 0.06mm to 6mm with 0.01mm difference. Below the lead also a perspex slice and water phantom were simulated. The layers were wide enough (5cm radius) so the scattering in lead layers and the backward scattering from water and perspex were included. Contribution to the dose of different parts of the geometry was determined to estimate which could be neglected. For example, the lead shield was found to have only small contribution to the total dose. A similar procedure was used to estimate the scattering from water layers. The first 2mm thin water layer just above the perspex made the greatest contribution. The absolute contribution faded with the water depth as expected. Our goal was to even out the dose distribution and not flux distribution, so a conversion from flux to dose had to be calculated.

2.3 Filter shape determination

In the initial calculation the dose distribution in water phantom without the filter was determined. On the basis of the parameters calculated with the above procedure the first filter shape was calculated. The iterative procedure was automated to calculate the optimal filter shape. Monte Carlo calculations were done by the MCNP4B [2] program. The correction to the filter shape in the previous iteration was based on comparing the calculated dose and required dose in the first water slice, considering the attenuation law and mean attenuation coefficient. In the next step the new (corrected) filter shape was used. The procedure was stopped when the differences between the last two filter shapes were negligible. This iterative procedure converged to the optimal filter shape very quickly because only few iterations were needed. However, because of the high Monte Carlo precision (<1%) even a few iterations took a long time. For this reason the filter calculation method was improved with the use of the depth dependent attenuation coefficient, scattering contribution and flux to dose conversion. In the scattering contribution it was assumed that the amount of in- and out-scattered photons was the same. This assumption turned out to be sufficient for our purpose.

3 RESULTS

3.1 Theoretical studies

The attenuation coefficient was calculated only for ^{192}Ir photon source [3] and on the required depth interval (from 0.06mm to 0.6mm) (Figure 3).

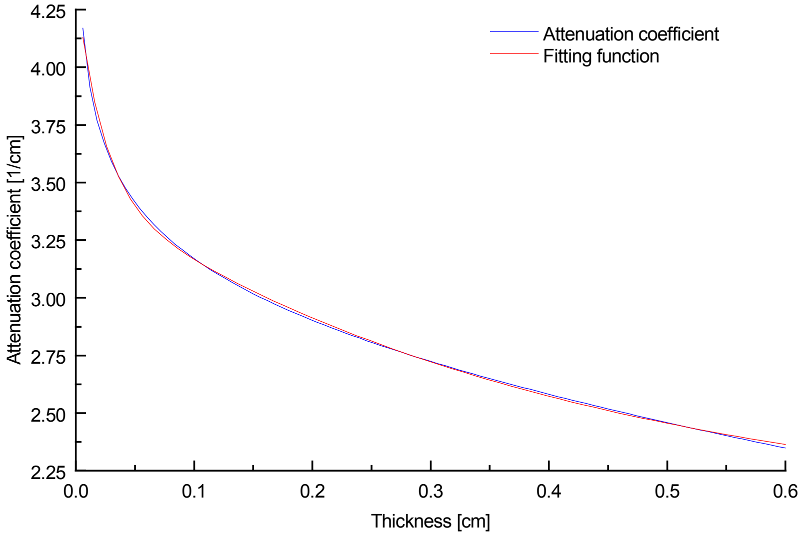


Figure 3 Dependence of the attenuation coefficient from the thickness of the lead layer (already including scattering and flux/dose ratio).

The scattering and transformation between the flux and dose are already accounted for in the result. The fitted function turned out to have the form of a second order exponential attenuation:

$$\mu = \mu_0 + A_1 e^{-\left(\frac{x-x_0}{t_1}\right)} + A_2 e^{-\left(\frac{x-x_0}{t_2}\right)} \tag{1}$$

For our case the parameters (μ_0=2.529 cm^{-1}, x_0=0 cm, A_1=0.641 cm^{-1}, t_1=0.056 cm, A_2=0.751 cm^{-1}, t_2=0.518 cm) were determined to fit to the numerical results. It turned out that the convergence is much slower when only the mean attenuation coefficient (3.1 cm^{-1}) [5] was used. In Figure 4 the integrated amount of the scattered particles in lead layers with different thickness is presented. The monodirectional source was directed perpendicular to the surface. The sum of all scattered particles in different distances from the source line was calculated.

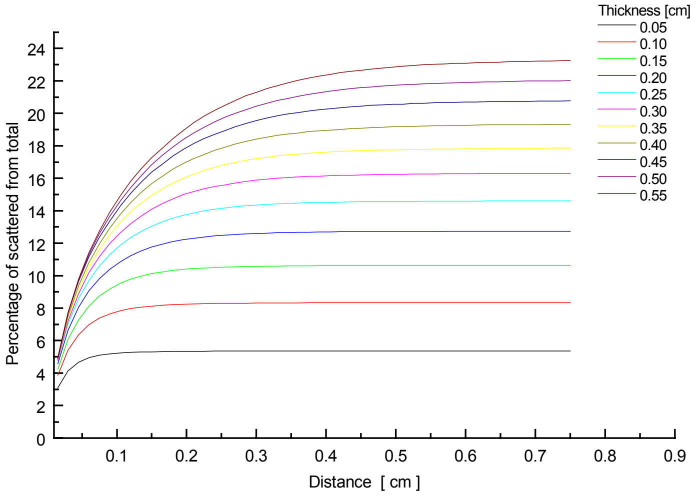


Figure 4 The sum of relative amount of scattered photons in dependence from thickness and source distance

The fitting function for the ratio between the flux and dose turned out to be linearly dependent on the thickness (Figure 5).

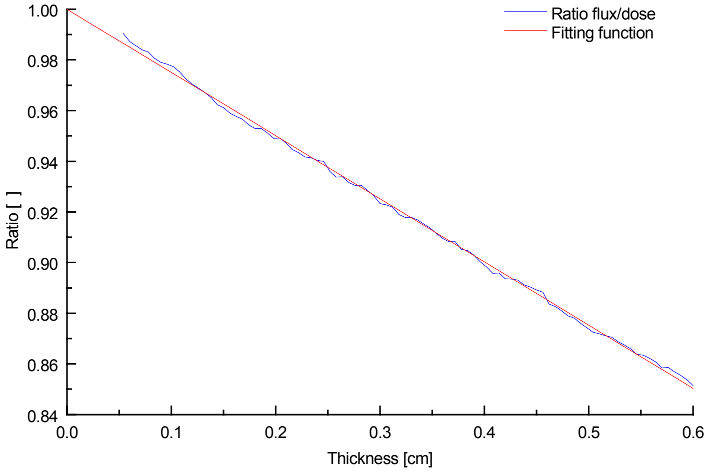


Figure 5 Dependence of the flux/dose ratio from thickness of the lead layer.

This relation can be described with the following fitting function:

Error! Objects cannot be created from editing field codes. (2)

In Equation (2) r represents the dimensionless ratio between flux and dose, x is the thickness and the fitting parameters are the following: a=1, b=0.249 cm^{-1}.

The summary of the contributions to the dose in the first water layer from the layers bellow as well as from other parts of the geometry is given in Table 2.

Table 2 The contribution of different part of water layers and lead shield. The numbers are percentage of the contributed dose to the total dose in the first water layer.

2nd layer	3rd layer	4th layer	5th layer	6th layer	7th layer
0.60	0.35	0.24	0.18	0.14	0.11
8th layer	**9th layer**	**10th layer**	**outside water**	**lead wall**	**lead roof**
0.09	0.07	0.05	0.31	0.25	0.009

3.2 Filter and dose results

In the geometry of the problem, the lead shield inner radius was 17.5 mm. In the inside area a cylindrically symmetrical filter with varied thickness was put. From the results it can be seen, that the filter thickness varied from 0mm (at the edge) to the maximum 3.37mm (Figure 6).

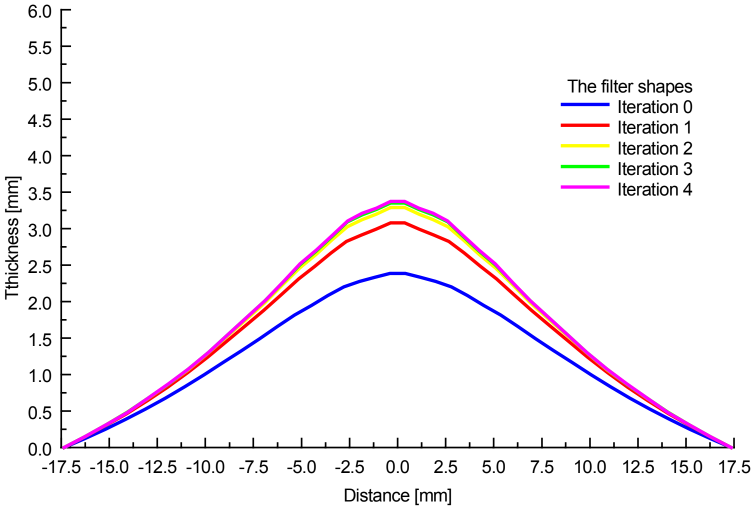


Figure 6 The filter shapes in different iteration. The difference between the last two iterations is almost unnoticeable.

The results presented here were obtained considering only the mean attenuation coefficient. The difference between the first two iterations is the cause of scattering and depth dependence of the attenuation coefficient, which were not considered. Taking this into account we achieved the optimal filter shape much quicker (Figure 7). Our goal, to even out the dose distribution, was achieved with extremely fast convergence in approximately only one third of the time spent previously.

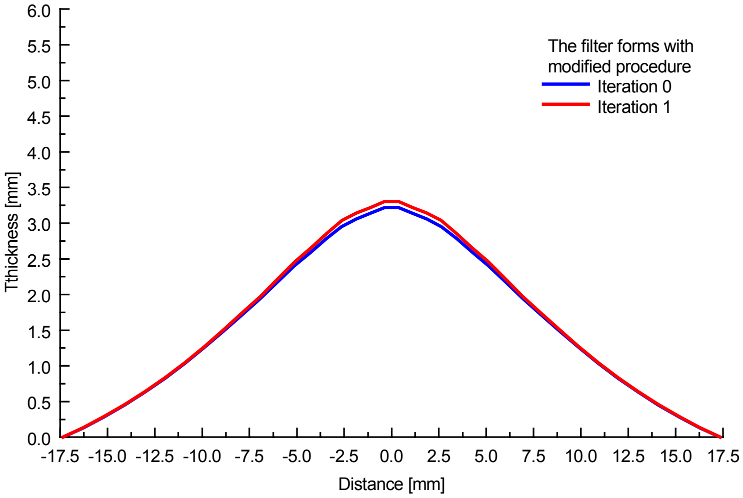
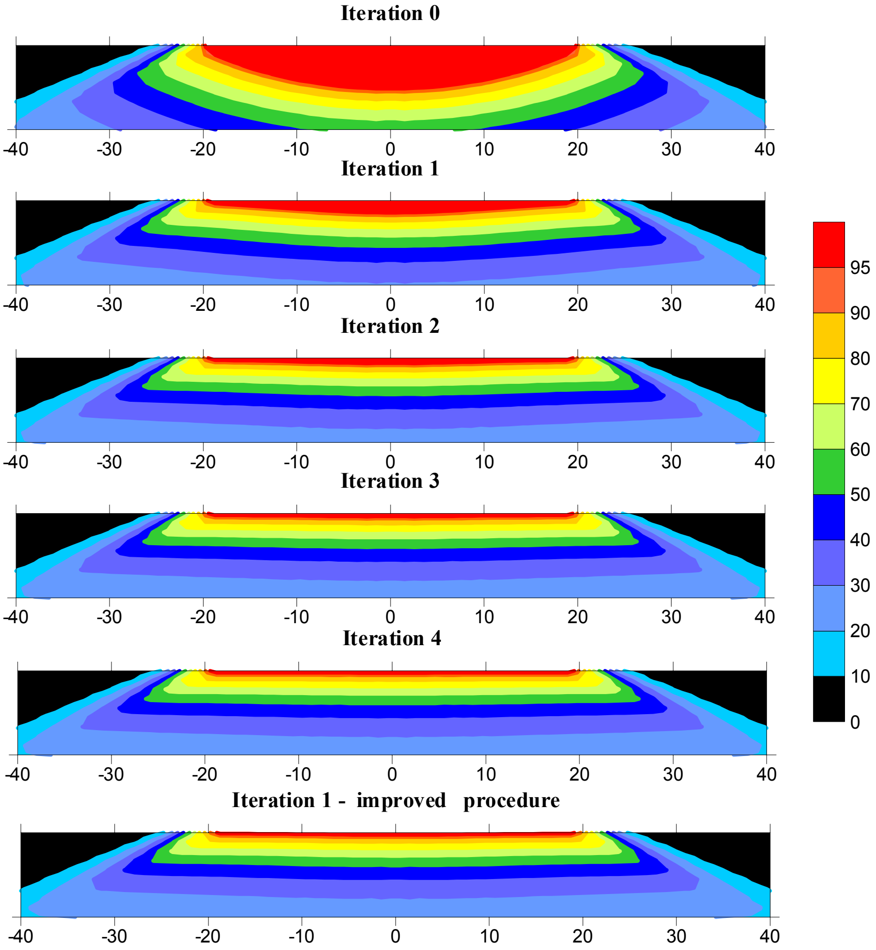


Figure 7 **The filter shapes for the improved procedure. Two iteration were enough for the optimal filter shape**

The significant improvement of the iterative procedure can also be seen from the dose distribution in the water phantom (bottom of Figure 8).


Iteration 0

Iteration 1

Iteration 2

Iteration 3

Iteration 4

Iteration 1 - improved procedure

Figure 8 **The dose distribution in water phantom for each iteration. At bottom is shown for the improved procedure. The radii are in mm while the total depth is 20mm**

With the use of our procedure the dose profiles in the water were smoothened to within a percent. The profiles (with and without optimal filter) were also experimentally verified. The experimental results and Monte Carlo calculations agreed very well. A similar procedure as ours could be applied for other cases with the same or any other brachytherapy source and different heights. Since in practice a high uniformity of the dose profile is usually required, these results can be very useful for skin cancer treatment.

4 CONCLUSIONS

The major goal of this paper was to explore the use of Monte Carlo simulation in skin radiotherapy treatment. The MCNP4B Monte Carlo code was used. The initially non-uniform dose distribution, which appears because of the proximity of the brachytherapy source to the surface, was evened out with the use of an appropriately shaped lead filter to within a percent. A highly efficient iterative procedure was used that enabled calculation of the filter shape in just one or two steps. The proposed procedure can be used for different heights and types of the source and for different filter materials.

5 REFERENCES

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