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# 1 Development and experimental validation of TPS software to determine the out- 2 of-field dose in radiotherapy beams

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## 6 I. INTRODUCTION

7 Even if the IMRT modality allows a more accurate definition of the target volume in radiotherapy, low doses are  
8 still delivered around the tumour to organs at risk. Epidemiological studies [1] demonstrated the relationship  
9 between peripheral doses and second cancers or heart diseases. Many experimental studies measuring the  
10 variation of the peripheral dose with the treatment parameters have been undertaken [2]. Up to now, very few  
11 studies have been performed on out-of-field Monte Carlo (MC) simulations. To our knowledge, *MCNPX* is the  
12 only one MC code that has been used to simulate the peripheral dose in radiotherapy [3]. Nowadays, there is thus  
13 no specific and accurate tool predicting the peripheral dose. Our aim is to develop a MC tool based on the  
14 *PENELOPE* code to compute the dose at the target volume and the organs at risk in order to enable a decrease of  
15 the peripheral dose by adapting the treatment's parameters. This tool will be implemented in a Treatment  
16 Planning System (TPS). In this paper we expose the first validation step of the out-of-field MC calculations,  
17 using a comparison with measurements in a specific large water tank.

## 18 II. MATERIALS AND METHODS

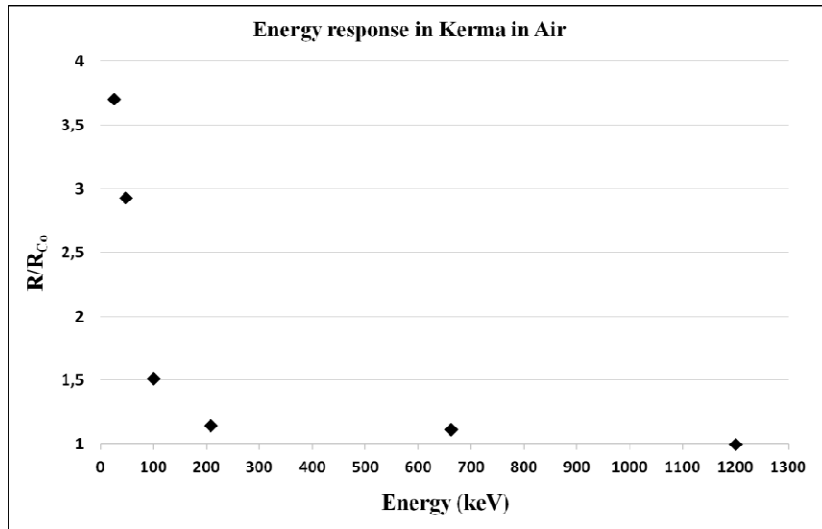
### 19 a. Experimental validation

#### 20 i. Irradiation configuration

21 The irradiations have been performed at LNH (French Primary Standard Laboratory) on a *GE Saturne 43* linear  
22 accelerator for 6, 12 and 20 MV beam qualities. We followed the configuration of the IAEA 398 protocol which  
23 implies the use of a  $10 \times 10$  cm<sup>2</sup> field, a skin surface distance (SSD) of 90 cm and a measurement at 10 cm depth  
24 in water. The only difference with the irradiation conditions proposed in this protocol is the use of a large water  
25 phantom (60 x 30 x 30 cm<sup>3</sup>) specially designed for this experiment. We considered a symmetric situation so only  
26 one side of the field has been explored.

#### 27 ii. OSL dosimeter

28 Another aim of this experiment is to validate the use of Al<sub>2</sub>O<sub>3</sub>:C detectors (*Nanodots*, Landauer) and the  
29 optically stimulated luminescence (OSL) technique for out-of-field dose measurements. The OSL technique  
30 presents advantages over the thermoluminescent technique (TL): no sample heating is required and the detector  
31 can be read several times. Moreover, OSL measurements are relatively cheaper than TL measurements [5].  
32 In this work, OSL dosimeters will be useful for the clinical validation of the code on an anthropomorphic  
33 phantom. Measurements performed with *Nanodots* have been compared to NE2571 ionization chambers (IC)  
34 measurements, considered as the reference data here.  
35



36

37 Fig. 1: Energy response of the OSL in Air Kerma. The ratio  $R/R_{C_o}$  corresponds to the raw response of the reader  
 38 divided by the raw response coming from the Cobalt

39

40 Four main factors are applied to the raw reading of the reader: particular sensitivity for each *Nanodot*, the  
 41 measured calibration factor, air calibration factor  $D_{water}/K_{air}$  correcting the fact that the calibration has been done  
 42 in air whereas the measurements have been performed in water and finally the energy dependence correction  
 43 factor.

44 Indeed, we measured a high over-estimation of the dose (by a factor 3 or 4) for photon belonging to a low energy  
 45 range ( $< 100$  keV) [Fig. 1]. Consequently, we developed an energy dependence correction protocol. It combines  
 46 the results of the experimental over-response curve with *PENELOPE* spectra calculations at measurement points.  
 47 The influence of this correction is described in the next part.

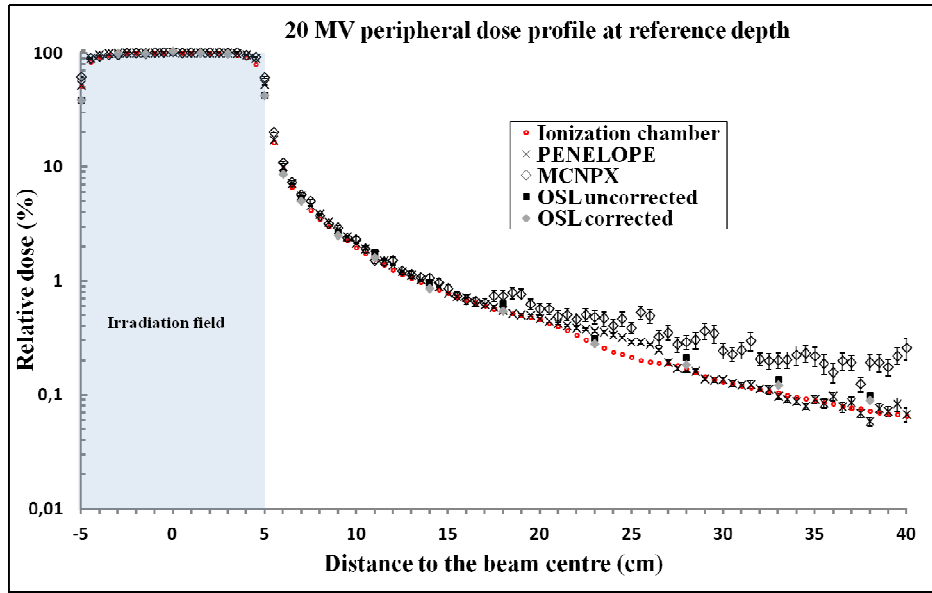
48 b. Calculation tool

49 The tool we are developing is based on the 2006 release of *PENELOPE* [4]. This code has been parallelized to  
 50 save computation time by running the calculations on a cluster of 372 processors. The simulations are run from  
 51 specific large Phase Space File (PSF) recorded in a  $60 \times 30$  cm<sup>2</sup> plane located at 90 cm from the source, after the  
 52 jaws of the modelled accelerator. An additional comparison has been done using the tally F6 of the *MCNPX* MC  
 53 code, the unique MC code used for this kind of studies until now. A specific feature of *MCNPX* is the  
 54 availability of the DXTRAN variance reduction method. It is specially adapted to the calculation far from the  
 55 primary beam and was tested in order to assess the gain to be expected on the out-of-field dose calculation  
 56 efficiency.

57 III. RESULTS

58 a. MC comparison

59 As results obtained for 6, 12 and 20 MV quality beams are quite similar, we have reported on Fig. 2 the  
 60 experimental and calculated out-of-field dose profiles obtained for 20 MV. These dose profiles have been  
 61 normalized to the maximum of dose.



62

63

Fig. 2: Experimental and calculated 20 MV dose profiles

64 As expected, PENELOPE and MCNPX results in the field are in good agreement with OSL and IC  
65 measurements (error < 3 %).

66 Out of the field, PENELOPE data are in good agreement with the IC measurements from the centre of the beam  
67 up to most distant points (40 cm). The statistical uncertainty associated to the most peripheral points is quite low  
68 (~15 %) regarding to the level of dose. The global mean error is about 3 % between these two sets of results.

69 Unlike the PENELOPE code, the MCNPX calculations over-estimate more and more the dose from 18 cm from  
70 the centre of the beam to the further positions. The global mean error is about 30 % between MCNPX and the IC  
71 values.

72 Consequently, compared to MCNPX, PENELOPE seems to be the best suited to simulate the out-of-field dose  
73 deposition with a high level of accuracy. To analyse the difference between the two MC codes, different cross-  
74 sections libraries were tested in MCNPX without significant improvement in the results. Discrepancies observed  
75 between both codes are still under investigation.

76

#### b. OSL comparison

77 On Fig. 2, one can observe that uncorrected OSLs always over-estimate the dose. Before applying the correction,  
78 the global mean error between OSL and IC measurements is around 5.6 %, after correction, it decreases to  
79 1.1 %. This result demonstrates the relevance of OSL dosimeters to measure low doses in out-of-field regions  
80 and thus to validate the calculations of the MC calculation tool in a real IMRT configuration with an  
81 anthropomorphic phantom.

82

#### c. DXTRAN tests

83 PENELOPE simulations conducted in this study are time consuming (more than 5 days of computation on 48  
84 processors for the whole water tank dose calculation). The code converges slowly for these calculations because  
85 of the low number of particles present out-of-field. Preliminary results obtained about the use of DXTRAN in  
86 MCNPX show an increase of the efficiency by a factor of 300. Therefore the implementation of such a tool in  
87 PENELOPE could significantly reduce the computation time of out-of-field dose and make enable the  
88 implementation of our tool in clinical practice.

89

## IV. CONCLUSIONS

90 The comparison of the MC tool based on PENELOPE with measurements gives satisfying agreements for out-  
91 of-field doses. The next step is the clinical validation with the OSL dosimeters within an anthropomorphic  
92 phantom and with an IMRT step-and-shoot treatment plan.

93 At the same time we are working on the acceleration of the calculations. The implementation of reduction  
94 variance techniques such as DXTRAN in the MC tool should be helpful by increasing calculation efficiency out  
95 of the beam. This implementation is undergoing and we will soon be able to give our first results of the  
96 acceleration of the *PENELOPE* code.

97 Part of the work has been done within the framework of EURADOS WG9.

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