Modeling an EPID to simulate radiotherapy transmission images

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(Dated: January 18, 2022)

Abstract: During an oncological treatment with external-beam radiotherapy, it is essential to avoid any movement or displacement of the patient. To guarantee this, modern medical accelerators are equipped with a device called EPID, capable of detecting the radiation that passed through the patient and create an image. The aim of this paper is to model a realistic geometry of an EPID to simulate radiotherapy transmission images and test if these changes in the considered geometry produce more useful and closer to reality dose distribution images. To do so, Monte Carlo simulations were done using the PENELOPE/penEasy code to simulate the transport of photons, and the images were analyzed employing the gamma index method.

I. INTRODUCTION

During an oncological treatment using radiotherapy it is crucial to be precise on two variables: the absorbed dose administered in order to eliminate the tumor and the millimetre position where the beam will fall upon, i.e. the exact position of the patient. Typically, kilovolt x-ray imaging is used to ensure the optimal position of the patient but, although it gives an image with enough information to correct any displacement, it can not be used to real-time monitor the position of the patient during all the radiotherapy sessions [1]. Since it is extremely important to detect and correct those movements immediately after they occur, there is need to find a real-time imaging method. Here is where the use of electronic portal imaging devices (EPIDs) takes place. These devices are attached to Linacs with the aim of obtaining and displaying images in a short period of time. To do so, it takes advantage of the photons that cross the patient, and that would be lost otherwise, to create an image from the dose distribution. The price to pay is that, since megavoltage photon beams are employed to treat the tumor –unlike the kilovoltage beams used in x-ray imaging, where the photoelectric effect is far more probable and so the quality of the image is better– there is a poor contrast between different tissues.

To understand and model an EPID helps to improve and make more successful treatments of radiotherapy thanks to the ability to correct any displacement of the patient. At the Hospital de Sant Pau (Barcelona), there is ongoing research trying to improve the simulations by modeling the geometry of an EPID in a simple way: a single layer of water as the detector of the radiation that went through the patient. This procedure lets us consider the EPID in our simulations and thus, we can generate an image of the patient using the dose distribution detected by this simulated device.

The next step is to build the geometry of an EPID in a more realistic way, according to the manufacturer specifications, with the aim of acquiring better results in our simulations [2]. To this end, the PENELOPE/penEasy Monte Carlo code was used to simulate photon-beam radiation images from different angles and compare the results using a realistic and simple geometries in a simulation with the measured dose distribution of a real process. To carry out the comparison, the gamma index method was adopted –with a Matlab function algorithm–, as it is currently used at the Hospital de Sant Pau.

II. MATERIALS AND METHODS

A. Monte Carlo simulation

We have employed PENELOPE/penEasy, a program widely used in medical physics, to simulate the transport of the photons, electrons and positrons emitted by a Linac, i.e. during radiotherapy treatments. PenEasy is a main program with multiple source models, tallies and paramaters to adjust the simulations to the case of study.

The activated tally was *Pixelated Imaging Detector* [3], which creates a pixelated image of the radiation hitting the detector. This tally requires a specific geometry built with planes (not quadratic equations) that, at least, form one box, which the program will regard as the detector. This single box would be the simplest geometry to consider: a single layer with the role of detector. The detector also has to be specified in the transport parameters section as a layer of a material that absorbs all the radiation that hits it. Our purpose is to make a more complex and realistic EPID adding some more bodies according to real EPIDs used in Linacs. PenEasy generates as many output files as activated tallies, hence in our case we will have a single output file with the dose in each pixel, which creates an image with anatomical information.

The files needed for the simulations, listed and detailed below, must be in the same folder of the penEasy executable during the simulation:

- *Phase-space file* (PSF): since the whole simulation of the transport of the photons of the Linac –from the W target until they reach the detector– would be time-consuming, we can skip the stage from the

target to the patient by using a file, the PSF, that contains all the information of their transport and that has been generated previously. We can do this because the properties of the beam, the transport parameters, the position of the collimators and the radiation field configuration is always the same until the beam reaches the patient. The PSF used has been provided by the Hospital de Sant Pau, and contains the output of weeks of simulation time. It would not be possible, considering the timing, to do the present study without them.

- Voxelized geometry (.vox): this file contains the anatomical information of the target, i.e. the portion of the body of the patient that will receive ionizing radiation.

- Geometry file (.geo): it corresponds to the geometry of the EPID and it is the main object of our work. We consider two types of geometry, a simple one and a realistic one to see how this improvement affects the simulation when it is compared with the measured –real– image. This part of the study is described in more detail below.

  Using penEasy, the geometry is handled with PENGEOM [4], a package that handles geometries defined with quadric surfaces. Here we adopted surfaces defined by the implicit equation (1) because the activated tally does not allow the use of quadratic equations, i.e. all the coefficients accompanying a quadratic term must be set as null, something that we would not be able to do with the reduced form of the quadrics.

$$\begin{aligned} F(r) = A_{xx}\,x^2 + A_{xy}\,xy + A_{xz}\,xz \\ +A_{yy}\,y^2 + A_{yz}\,yz + A_{zz}\,z^2 \\ +A_x\,x + A_y\,y + A_z\,z + A_0 = 0. \end{aligned} \tag{1}$$

  The geometry file contains the coefficients that define each surface, which can then be rescaled, rotated and/or shifted.

- Material files (.mat): we must provide the cross sections and additional information of the materials that are defined in both the voxelized and quadric geometries. These files are generated with the `material.exe` preprocessor of PENELOPE, entering the mass density and composition of each material. For the voxelized one we will be using adipose tissue, bone, lung and muscle; while in the quadric geometry we use the EPID's materials: copper, amorphous silicon, glass and gadolinium oxysulfide.

- penEasy.in: file with all of the parameters and specifications required by penEasy and PENELOPE. It contains general information like the number of histories to be simulated, the initial random seeds, the alloted time, etc, the specification of the radiation source, the name(s) of the geometry file(s), the values of the transport parameters of each material and, finally, which tallies will be turned on or off. The transport parameters where set as follows:

  – Absorption energies (E_{abs}) equal to 500 keV for electrons/positrons and 10 keV for photons, except for the detector material, which was set to 500 keV as well.

  – The cutoff energy for photons (Wcc) was set to $1keV$,

  – The cutoff energy for radiation (Wcr) was set to 10 keV.

  In this study, only the *Pixelated Imaging Detector* tally has been turned ON. PenEasy.in contains the information that this tally receives: the detector material, the properties of the pixel –set to a size of 0.1×0.1 cm–, the threshold energies, the requested uncertanty –set to zero–, among others.

- penEasy.exe: executable file, generated with a fortran compiler. By the date of this project, it was the latest version with a modification that allows the use of more than ten materials.

In the present investigation, we run simulations for the simple and the realistic geometries at 7 gantry angles, i.e. 14 simulations. This lets us have a good idea of the accuracy of the results and make sure that the improvement caused by the realistic geometry is not fortuitious. We repeated all 14 simulations three times so as to generate the images with smaller statistical uncertainties.

To understand the position of the patient regarding the Linac and the EPID and the use of a PSF, Fig. 1 represents a patient laying during the radiotherapy treatment. The structure above is the upper part of the Linac (s1) –from the W target to the filters– and the collimators (s2). We can see that the EPID is not represented, but it would be located underneath the patient, perpendicular to the photon beam. The complete structure, including the EPID, which is always perpendicular to the photon beam, is able to rotate around the patient to irradiate the tumor from the desired directions. This is important because we use a PSF that includes steps s1 and s2, as in each direction the collimators and jaws should have different apertures, in each simulation with a different angle, the PSF will also be different: 7 PSFs will be considered. Finally, s3 represents the patient and it is not considered in the PSF we are using, although it will be always the same in our simulations, not considering the PSF of the s3 region give us the option to remove the voxelized geometry if there is interest in it.

B. Realistic EPID geometry

A realistic EPID, following the instructions of the manufacturer [5], can be understood as a block of four layers

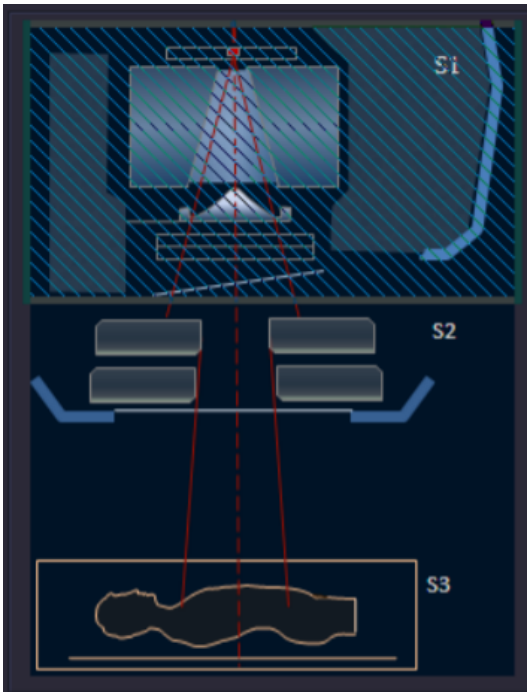

FIG. 1: Representation of the Linac, the patient and the photon beam. Image extracted from Primo User's Manual [6].

of different materials wrapped in a plastic shell, as shown in Fig. 2.

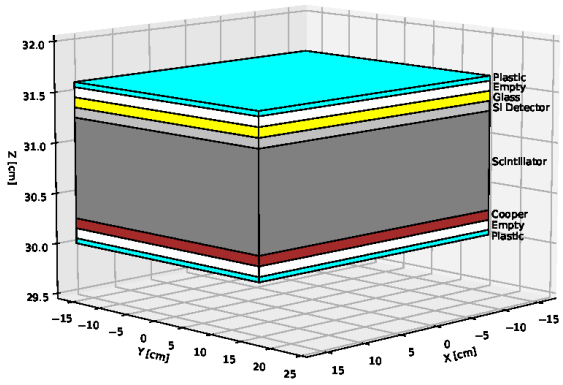

FIG. 2: Geometry used in the simulations to model the realistic EPID.

Notice that the radiation beam comes from the bottom, hence the order in which the layers are crossed is as follows: first the slab of copper, then the Gd_2O_2S scintillator, next the detector, which is a layer of amorphous silicon, and finally a thin layer of glass, here represented by SiO_2. As the four layers are composed of well-known substances, they can be considered accurate. The only layer that can suggest some kind of controversia is the scintillator as it is defined as a homogeneous material but it contains the electronics of the device and so it is composed of a grid of crystallographic cells. All these materials are located inside a plastic case, as we can see in the figure as the top and the bottom layers.

It is worth mentioning that the active tally does not allow any type of implicit geometry or any kind of reduction in the definition of geometry. Thereby, every layer has to be defined as a single box, and therefore each plane has to be defined as many times as it is used.

C. Image Analysis

Once we have the simulation results, a comparison method is needed to assess whether the realistic geometry has improved the model and there is a visible difference or not. At the end of the simulations we have three kinds of image according to their dose distributions: the reference image of the patient (measured), the image we have acquired with the simulation done with the simplest geometry (only one layer representing the EPID) and the image we have acquired with the realistic geometry we have built. In order to get the comparison between this images we need our last tool: the γ index [7].

Gamma analysis is an algorithm that takes as an input the simulated and the measured absorbed dose distributions and the range of error we accept both for the absorbed dose and the position (the clinical standards usually are around 3% for the dose and 3 mm for the position). With these inputs, the algorithm calculates the distance between the points of the two dose distributions (the measured and simulated ones) superimposing the isodose distributions. Then, an evaluation is made of whether the distance is inside our range of acceptance or not and it gives a relative value, called γ index, which is less than or equal to 1 if it is inside our range and greater if it has failed the test and so its error is greater than our acceptance range. In other words, if $\gamma = 1$ we will have an error of precisely our dose acceptance value and the position acceptance value, if $\gamma > 1$, the error is greater and if $\gamma < 1$, the error is lower. Once the analysis is made, we can get the mean value of all the gamma index of each pixel of the image to get an idea of how close is the simulated dose distribution to the real one (the measured one). Note that the lower the mean, the closer is our calculated dose distribution to the measured one. In our study, the hypothesis is that, when conducting this analysis for the simple geometry and then again for the realistic one, the realistic geometry dose distribution will be closer to the reference dose distribution, i.e. the mean value of gamma index of the realistic geometry will be lower than the mean of the gamma index considering the simple geometry.

The gamma index has been adjusted to a dose acceptance value of 7% and a position acceptance value of 3 mm. This deviation from the standard clinical values will only affect in those regions where the result is far from reality in a way that the gamma index will saturate in a threshold value: we will not know the exact gamma index from a value on, but we will know it is far from one and so, it is not adjusted with reality. We do so to

improve the computational process and get the results in a reasonable time. To implement the gamma analysis, we used an algorithm on Matlab, after normalizing the pixel gamma index values in a range of $[0, 1.1]$.

III. RESULTS AND DISCUSSION

After the simulations, we have two data files (one using the simplest geometry and another using the realistic geometry) for the 7 angles. The simulations and the measurements have been executed three times in order to enrich the research and avoid random and instrumental errors. Each of the images considered is, actually, the mean of three dose distributions. After the simulations, the gamma analysis have been made on the resulting data files to compare how close the result is to reality.

The gamma analysis is given as a grid of the gamma index of each pixel. Making the average of all gamma index values we get an idea of how each geometry adjusts to reality. The results of our simulations are shown in Table I. The lower the mean gamma index, the closer is the simulated image to the real (measured) one.

TABLE I: Mean of gamma index for each dose distribution.

Angle (^{o})	γ, simple geom.	γ, realistic geom.
60	0.2312	0.1738
100	0.2134	0.1179
140	0.1847	0.1489
180	0.2133	0.2290
220	0.2071	0.1608
260	0.1825	0.1681
300	0.1567	0.1554

As we can see, with the parameters chosen in the gamma analysis, six out of seven simulated dose distributions with the realistic geometry were closer to the measured dose than those using the simplest geometry. Therefore, the improvements made in the geometry had a visible effect in the simulation.

Furthermore, making use of visualization tools we can create an image of the gamma index, so it will represent those pixels which misalign the real dose distribution. In the figures below we show the images of the gamma index for the case of 60^{o} and 100^{o}, Figs. 3 and 4 represent the gamma index using the simple geometry (top) and the improved and realistic geometry (bottom).

From these figures it seems clear that the realistic geometry improves the dose distribution image (the amount of red is reduced), but we can also observe that those regions where the dose distribution is further from reality (more red) in the first image, does not improve in the second image. The reason for this fact is that, with improving the geometry, we are only improving the detection of the radiation coming from the layers that compose the EPID, but not all the other factors that can affect the

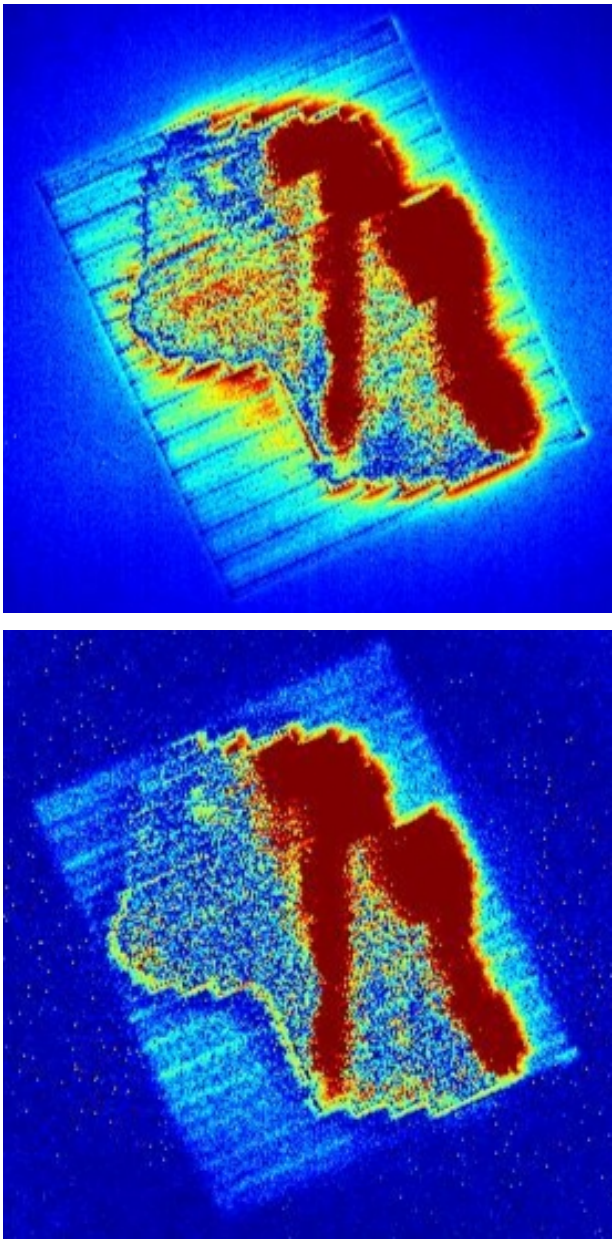

FIG. 3: γ analysis for the dose distributions of the simple (top) and realistic (bottom) geometries and an angle of 60^{o}.

results such as secondary photon beams, the transport parameters used or other processes.

IV. CONCLUSIONS

Monitoring the misalignment of the patient during a radiotherapy treatment is crucial to guarantee the success of the treatment. Taking advantage of the radiation used to eliminate the tumor by means of an EPID seems like a useful method to do so. In order to improve the Monte Carlo simulation results considering the EPIDs we have seen how much better can the resulting dose distri-

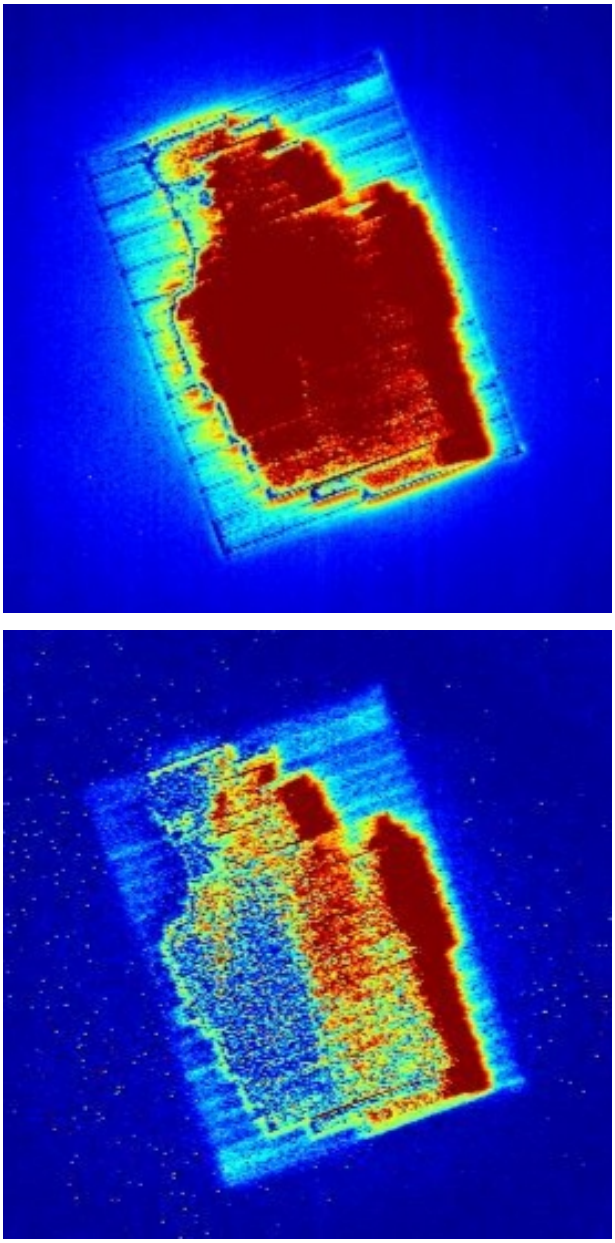

FIG. 4: γ analysis for the dose distributions of the simple (top) and realistic (bottom) geometries and an angle of 100^{o}.

butions be if we consider a realistic geometry of an EPID as compared to a simple geometry.

In the study we have used the code PENELOPE/penEasy to carry out the Monte Carlo simulations, and the gamma index analysis to compare and establish a qualitative and quantitative measurement of the accuracy of our dose distributions to reality.

The results have concluded that the realistic geometry seems to improve the simulations to a certain extent: the detected radiation coming from the multiple layers of the EPID are accurate enough in comparison with the simple geometry, but those regions where the photons are created by other processes did not improve with the realistic geometry. Although the use of a realistic geometry gives us the possibility to improve the simulation results, these still have deviations to correct. The simulation misalignment is not coming only from the geometry. Therefore, a study of the secondary photons reaching the detector or other processes that can interfere in the simulation should be carried out.

Acknowledgments

I would like to thank my tutors José M. Fernández-Varea and Artur Latorre for their help and guidance; it has been a pleasure to learn from them. I would also like to acknowledge my family for their inconditional support.

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