Optimizing portal dose calculation for an amorphous silicon detector using Swiss Monte Carlo Plan

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Purpose: Modern treatment planning systems (TPS) are able to calculate doses within the patient for numerous delivery techniques as e. g. intensity modulated radiation therapy (IMRT). Even dose predictions to an electronic portal image device (EPID) are available in some TPS, but with limitations in accuracy. With the steadily increasing number of facilities using EPIDs for pre-treatment and treatment verification, the desire of calculating accurate EPID dose distributions is growing. A solution for this problem is the use of Monte Carlo (MC) methods. Aims of this study were firstly to implement geometries of an amorphous silicon based EPID with varying levels of geometry complexity. Secondly to analyze the differences between simulation results and measurements for each geometry. Thirdly, to compare different transport algorithms within all EPID geometries in a flexible C++ MC environment.

Materials and Methods: In this work three geometry sets, representing the EPID, are implemented and investigated. To gain flexibility in the MC environment geometry and particle transport code are independent. That allows the user to select between the transport algorithms EGSnrc, VMC++ and PIN (an in-house developed transport code) while using one of the implemented geometries of the EPID. For all implemented EPID geometries dose distributions were calculated for 6 MV and 15 MV beams using different transport algorithms and are then compared with measurements.

Results: A very simple geometry, consisting of a water slab, is not capable to reproduce measurements, whereas 8 material layers perform well. The more layers with different materials are used, the longer last the calculations. EGSnrc and VMC++ lead to dosimetrically equal results. Gamma analysis between calculated and measured EPID dose distributions, using a dose difference criterion of ± 3% and a distance to agreement criterion of ± 3 mm, revealed a gamma value < 1 within more than 95% of all pixels, that have a higher dose than 3% of the measured maximum dose.

Conclusions: The same geometry can be used to compare transport codes within the flexible C++ MC environment. A simplified 8 layer geometry results in similar EPID dose distributions compared with the accurate 24 layer geometry by gaining about a factor of 6 in CPU-time. The implementation of the amorphous silicon detector in our MC system has the potential to perform independent pre-treatment as well as treatment verification.
1. Introduction
In current radiotherapy more complex and computer driven techniques as e.g. intensity modulated radiation therapy (IMRT) are used. Electronic portal image devices (EPIDs) that are mounted directly on a linear accelerator, have been well established for patient setup and IMRT verification purposes [1]. For a few years EPIDs based on amorphous silicon have been manufactured. Digital images are immediately available after an exposure and can be interpreted by medical personnel at the linear accelerator. At the beginning of a treatment session the position of the patient is verified only with a few centigray [2][3][4][5].

Apart from patient setup, EPIDs are used for IMRT pre-treatment quality assurance measurements to check dose distributions calculated by a treatment planning system (TPS) and the dose delivery before a treatment [6]. For this purpose the dose distribution is recalculated in a water phantom at a depth defined by a protocol. The EPID signal is then calibrated to this water depth. Resulting dose distributions were compared using analysis as e.g difference plot or a distance to agreement criterion. Consequently, an independent dose check is achieved. Additionally, the dose below the patient is of great interest to check treatment delivery in each fraction. Portal dose image predictions are available in some TPS, but are limited in accuracy [7]. A solution for this problem of calculating EPID dose distributions is the use of Monte Carlo (MC) methods, which has already been shown by several groups [8][9][10][11][12][13][14]. Involving MC methods have the major drawback of long computation times [15]. The aim of this work is to implement EPID geometries that vary in layer complexity in a given flexible MC environment. By this means the geometry complexity needed for a given accuracy is investigated. Moreover, EPID dose distributions calculated by using the available transport algorithms in the MC environment within the same geometry are compared with measurements. The dosimetric accuracy and the performance of each transport algorithm are assessed. Based on these investigations the complexity of geometry and transport algorithm to calculate EPID dose distribution in the MC environment are optimized.

2. Methods and Materials

2.1. EPID

All measurements are performed with the Portal Imagery aS500 on a Clinac 2300 C/D with a 80 leaf multi leaf collimator (MLC) from Varian Medical Systems, Palo Alto, USA. Measured pixel signals at location (x, y) were darkfield (DF) and flood-field (FF) corrected according to equation (1):

\[
\text{corrected signal} (x, y) = \frac{\text{pixel signal} (x, y) - \text{DF} (x, y)}{\text{FF} (x, y) - \text{DF} (x, y)}
\]  

(1)

This signal can then be related to the linear accelerator Monitor Unit (MU) output. Thereby the measured pixel signal at different source to surface distances (SSD) can be linear related to the dose delivered (in MU) by taking the inverse square law into account. Further measurements using different dose rates resulted in a linear relationship, showing the independency of the pixel signal on dose rate [12][14]. A detailed description of the signal forming process including optical transports and reading electronics within the EPID can be found in [14][16][17][18]. The amorphous silicon detector consists of three major parts as drawn in figure 1. In the copper plate a part of the incident photons are

![Figure 1: Schematic drawing of the amorphous silicon detector with main components: copper plate, gadoliniumoxysulfate (GOS) slab and amorphous silicon (a-Si) plate with embedded electronics for readout](image-url)
converted into high energetic electrons. These electrons deposit some of their energy in the gadoliniumoxysulfate (GOS) slab. The energy deposition is converted into optical photons which are detected from photodiodes in the amorphous silicon (a-Si) layer. To calibrate the measured signal to a dose value following protocol has been used [19]: the mean of the central 11x11 pixel values of a 10x10 cm$^2$ open field measurement is set to the value of an ion chamber central axis measurement of the same 10x10 cm$^2$ open field at a water depth of 7.35 cm resulting in one conversion factor applied to all signal values. In this protocol the EPID is located at SDD = 140 cm, whereas the ionization chamber measures the dose at isocenter within a solid water phantom.

2.2. MC environment

In this work the Swiss Monte Carlo Plan (SMCP) is used as the MC environment [20]. An advantage of this environment compared with others is the ability to define a geometry independently from a transport algorithm. Consequently, the same geometry is reused to calculate dose distributions with different transport algorithms. In this work three algorithms are used, EGSnrc [21] [22], VMC++ [23][24][25][26] and PIN an in-house developed algorithm based on EGSnrc. PIN simplifies multi-scattered electrons by using a mean pathlength for interactions. Empirical cross section data are taken from the NIST program XCOM. The simulation of interaction events is done as in EGSnrc with IBCMP = 0. A scheme of the SMCP is shown in figure 2.

Figure 2: Scheme of main elements of SMCP as beam model, beam modifiers, dose calculation within the patient and implemented EPID-geometry

A beam model is used to simulate the LINAC treatment head. The beam modifiers as e.g. the MLC, wedges or blocks are simulated for each treatment field. The acquired CT-images are imported as DICOM data set and CT numbers are converted into electron densities. To compute dose within the patient, a dose engine is used. All particles leaving the patient geometry are reused in the new implemented geometries (see subsection 2.3) to calculate additionally an EPID image. In each part of a simulation the user can chose different transport algorithms [27].
2.3. Implemented EPID geometries
In this work three different geometries are implemented in the SMCP environment. A geometry that accurately describes the used Portal Imager aS500 consisting of 24 slabs and 14 different materials based on the information of the vendor (extended geometry) [8], a simplified 8 slab geometry with 6 materials and a 2 cm thick water slab. These geometries are similar to those presented in [15]. To implement these geometries different classes of the SMCP environment are used.

The measured EPID image can be calculated from the energy deposition within the thin GOS layer as shown by several groups [8][10][11][12][13][14][16][17][18].

To calibrate the calculated EPID dose distributions the mean of the central 11x11 pixel values of the 10x10 cm² open field simulation at SDD = 140 cm is set to the value of an ion chamber central axis measurement of the same 10x10 cm² open field at a water depth of 7.35 cm. This is similar to the measurement calibration. Consequently, nine calibration factors were determined, one factor for each geometry calculated with one transport algorithm (3x3).

2.4. Simulations
Two different kinds of simulations were performed. Firstly, the implemented geometries are benchmarked using a 10x10 cm² field. Secondly, EPID dose distributions of IMRT fields without a patient were simulated. In all simulations EGSnrc for the linac head and VMC++ for the beam modifiers were used. To compare the different transport algorithms, only the transport code within the EPID geometry was changed. A more detailed description is given in the following subsections. Dose deposition within the GOS layer and CPU-time consumptions are determined for these simulations. Resulting dose distributions for the EPID were then compared with corresponding measurements with the gamma analysis introduced by Low [28]. For all cases the criteria were set to 3% difference of maximum measured dose and a distance to agreement of 3 mm. All calculations were performed on a PC with an AMD Athlon 64 3400+ 2.4 GHz processor.

2.4.1. Benchmarking with 10x10 cm² beams
6 MV and 15 MV 10x10 cm² open field beams were simulated at SDD = 140 cm at gantry and collimator angle of 0°. These simulations were first checked for the correct set up and then compared with measurements. Simulations with 1e9 particles impinging on the EPID were performed and the deposited energy within the EPID GOS layer was scored leading to an uncertainty of 2.7% for dose points higher than 50% of the maximum dose.

2.4.2. IMRT-field simulation
The EPID dose distribution for a clinical IMRT field is simulated. For this purpose, the MLC controlling file is used generated by the leaf motion calculator in the TPS (Eclipse, Varian Medical Systems, Palo Alto, USA). A dose distribution using 1e8 particles impinging on the EPID were performed and the deposited energy within the EPID GOS layer was scored leading to an uncertainty of 2.3% of pixel doses above 50% of the maximum dose. It is compared with a measurement at SDD = 140 cm following our pre-treatment verification protocol [19]. Computed and measured IMRT-field dose distributions were calibrated and a gamma analysis was performed subsequently.

3. Results
3.1. Benchmarking with 10x10 cm² beams
To check the field size profiles of the simulated 6 MV 10x10 cm² open field are shown in figure 3. As expected at SSD = 140 cm the profiles fall off at 7 cm off-axis for the 10x10 cm² open field for every implemented geometry and transport algorithm. All profiles are normalized to the mean value of the central 11x11 pixel values. Additionally the measured profile is added. Similar results for the profiles of the 10x10 cm² open field at 15 MV are obtained.
Figure 3: Zoomed profiles of the simulated 6 MV 10x10 cm$^2$ open field for each geometry. Results of each transport algorithm are compared with the measurement in the different panels.

In table 1, calculated mean dose values of the 6 MV 10x10 cm$^2$ open field from the central 11x11 pixel values are given. In table 2, differences of the mean values with respect to EGSnrc are given. Obviously, EGSnrc and VMC++ are dosimetrically equivalent, whereas PIN shows smaller dose responses. Similar results are obtained for the 15 MV beam as summarized in tables 3 and 4.
Table 1: mean dose values of central 11x11 pixel in the geometries coupled with three transport algorithms

| E = 6 MV       | EGSnrc  | PIN   | VMC++   |
|----------------|---------|-------|---------|
| PVI_water      | 6.69E-15| 6.31E-15| 6.71E-15|
| PVI_8slab      | 7.92E-15| 5.88E-15| 7.74E-15|
| PVI_24slab     | 7.24E-15| 4.74E-15| 7.18E-15|

Table 2: differences of mean dose values with respect to EGSnrc

| E = 6 MV       | (VMC-EGS)/EGS | (PIN-EGS)/EGS |
|----------------|----------------|----------------|
| PVI_water      | 0.3%           | -5.6%          |
| PVI_8slab      | -2.2%          | -25.7%         |
| PVI_24slab     | -0.8%          | -34.5%         |

Table 3: mean dose values of central 11x11 pixel in the geometries coupled with three transport algorithms

| E = 15 MV      | EGSnrc  | PIN   | VMC++   |
|----------------|---------|-------|---------|
| PVI_water      | 3.33E-14| 3.08E-14| 3.39E-14|
| PVI_8slab      | 4.37E-14| 3.26E-14| 4.09E-14|
| PVI_24slab     | 3.68E-14| 2.46E-14| 3.63E-14|

Table 4: differences of mean dose values with respect to EGSnrc

| E = 15 MV      | (VMC-EGS)/EGS | (PIN-EGS)/EGS |
|----------------|----------------|----------------|
| PVI_water      | 1.8%           | -7.7%          |
| PVI_8slab      | -6.3%          | -25.4%         |
| PVI_24slab     | -1.3%          | -33.3%         |

Finally, a gamma analysis in the EPID plane (SDD = 140 cm) for all simulations was done. The corresponding measurement was taken as reference data. Measurements and simulated dose distributions were calibrated and compared with each other. As an example, the gamma distribution resulting of such an analysis using the 24 slab geometry and EGSnrc as transport algorithm is shown in figure 4. 98.7% of all pixel with a value higher than 3% of the maximum measured dose result in a gamma value < 1.

Figure 4: Gamma evaluation result for a 6 MV 10x10 cm² field at SDD = 140 cm is shown. The simulation was done with the 24 slab geometry using EGSnrc. The calculated dose distribution results in a gamma value < 1 for 99.1% of all pixel with a value higher than 3% of the maximum measured dose.

Table 5 summarizes the gamma evaluation analysis for the 10x10 cm² open field beams at 6 MV and 15 MV, respectively. Values represent the relative number of pixels with a dose higher than 3% of the maximum dose where the gamma value is < 1. Values in brackets show the result for the 15 MV open field beams. Obviously, the water slab geometry is not able to reproduce the measurements within the given tolerances. On the other hand, the simplified 8 slab geometry and the 24 slab geometry are more or less equivalent.
Table 5: Summary of gamma evaluation analysis for 6 MV (15 MV) 10x10 cm\(^2\) fields. Values are percentages of pixels with a dose higher than 3\% of the maximum measured dose where the gamma value is < 1.

| transport algorithm | water   | 8 slab   | 24 slab  |
|---------------------|---------|----------|----------|
| EGS\textsc{src}     | 96.3 (97.9) | 99.8 (99.7) | 99.1 (99.9) |
| PIN                 | 98.6 (99.7) | 99.9 (99.9) | 99.6 (99.9) |
| VMC++               | 95.8 (99.1) | 99.6 (99.2) | 99.6 (99.9) |

The CPU-time requirements for the simulations of 1e9 incident particles are shown in figure 5. It is obvious that the water slab is the fastest whereas the 24 slab geometry requires the longest time. In these examples of 10x10 cm\(^2\) beams the CPU-time for a simulation with the water geometry is about 30 min, 25 h with the 24 slab geometry and 4 h with the simplified 8 slab geometry. Only a neglectable CPU-time requirement difference between 6 MV and 15 MV is seen.

3.2. IMRT-field simulation
As an example in figure 6 the result of a simulated head and neck field applied with the sliding window technique is shown. The simulation was calculated with the 8 slab geometry using VMC++. Both dose distributions are drawn in grey levels as percentage of the maximum dose of the measurement. In 6 c) the result of the gamma analysis is presented. 99.9\% of all simulated pixel with values higher than 3\% of the maximum dose have a gamma value < 1.
4. Discussion
In this work, geometries, representing an amorphous silicon detector, in different levels of complexity were implemented and investigated. Reducing complexity while maintaining accuracy of an 24 slab geometry lead to a significant decreased CPU-time. A geometry build by 8 layers was shown to reproduce results of the 24 slab geometry with respect to the involved analysis criteria. The dosimetric behaviour and required CPU-time requirement of three different transport codes within the same geometry was investigated. Results showed that PIN needs further investigations to retrieve answers concerning differences of dose deposition within the EPID geometry. EGSnrc and VMC++ showed no dosimetric differences but in CPU-time requirement. Consequently, in a clinical use, the implementation of the 8 slab module together with potential variance reduction techniques as e.g. Russian roulette, photon splitting or recycling [26] should be suitable. However, further investigations have to be performed regarding variance reduction techniques to speed up the calculation. By using the SMCP code an independent dose calculation is available which results in a patient and EPID dose...
distribution within one simulation. The possibility of calculating treatment and pre-treatment IMRT fields with and without the patient, respectively, allows to verify treatment delivery in-vivo.

5. Conclusion
The same geometry can be used to compare different transport codes within the flexible SMCP environment. A simplified 8 slab geometry results in similar EPID dose distributions compared with the 24 slab geometry by gaining about a factor 6 in time. The implementation of an amorphous silicon detector in the SMCP environment has the potential to perform dosimetric pre-treatment as well as treatment verification.

References
[1] Palta J R and Mackie T R 2003 Intensity-Modulated Radiation Therapy—The State of the Art, Medical Physics Publishing Madison WI 888
[2] Pisani L, Lockman D, Jaffray D, Yan D, Martinez A and Wong J 2000 Setup error in radiotherapy: on-line correction using electronic kilovoltage and megavoltage radiographs Int. J. Radiat. Oncol. Biol. Phys. 47 825-39
[3] Vetterli D, Riem H, Aebersold D M, Greiner R H, Manser P, Cossmann P, Kemmerling L, Born E J and Mini R 2004 Introduction of a novel dose saving acquisition mode for the PortalVision aS500 EPID to facilitate on-line patient setup verification Med. Phys. 31 828-31
[4] Vetterli D, Thalmann S, Behrensmeier F, Kemmerling L, Born E J, Mini R, Greiner R H and Aebersold D M 2006 Daily organ tracking in intensity-modulated radiotherapy of prostate cancer using an electronic portal imaging device with a dose saving acquisition mode Radiother. Oncol. 79 101-8
[5] Antonuk L E 2002 Electronic portal imaging devices: a review and historical perspective of contemporary technologies and research Phys. Med. Biol. 47 R31-R65
[6] Van Esch A, Depuydt T and Huyskens D P 2004 The use of an aSi-based EPID for routine absolute dosimetric pre-treatment verification of dynamic IMRT fields Radiother Oncol. 71 223–34
[7] McCurdy B M and Pistorius S 2000 A two-step algorithm for predicting portal dose images in arbitrary detectors Med. Phys. 27 2109-16
[8] Siebers J V, Kim J O, Ko L, Keall P J and Mohan R 2004 Monte Carlo computation of dosimetric amorphous silicon electronic portal images Med. Phys. 31 2135-46
[9] Steciw S, Warkentin B, Rathee S and Fallone B G 2005 Three dimensional IMRT verification with a flat-panel EPID Med. Phys. 32 600-12
[10] Monajemi T T, Steciw S, Fallone B G and Rathee S 2004 Modeling scintillator-photodiodes as detectors for megavoltage CT Med. Phys. 31 1225-34
[11] Jarry G and Verhaegen F 2005 Electron beam treatment verification using measured and Monte Carlo predicted portal images Med. Phys. 30 4977-94
[12] McCurdy B M Luchka K and Pistorius S 2001 Dosimetric investigation and portal image prediction using an amorphous silicon electronic portal imaging device Med. Phys. 28 911-24
[13] Bissonnette J P, Munro P and Cunningham I A 2003 Monte Carlo simulation of the image formation process in portal imaging Med. Phys. 30 3243-50
[14] Manser P 2003 Verifikation von intensitätsmodulierter radiotherapie mit einem amorphen silizium portal image device Thesis ETH Zurich Switzerland
[15] Chin P W 2005 Monte Carlo portal dosimetry Thesis University of Wales
[16] Treier R 2002 Dynamisches Verhalten eines Portalbild Detektors Diploma ETH Zurich Switzerland
[17] Barengo R N 2002 Monte Carlo Simulationen eines Portalbild Detektors Diploma ETH Zurich Switzerland
[18] Bühlmann R 2003 Portale Dosis in der intensitätsmodulierten Radiotherapie Diploma ETH Zurich Switzerland

[19] Vetterli D, Born E J, Cossmann P, Mini R, Manser P and Rüeggsegger P 2002 IMRT verification in clinical routine using portal imaging 44th AAPM Meeting, Montreal (poster)

[20] Figini S P 2002 Monte Carlo treatment planning for photon beams: Swiss Monte Carlo Plan Med. Phys. 30 1516 (abstract)

[21] Berger M J 1963 Monte Carlo Calculation of the penetration and diffusion of fast charged particles Alder B Fernbach S and Rotenberg M Meth. in Comput. Phys. Academic New York Vol. 1 pp. 135–215

[22] Bielajew A F, Hirayama H, Nelson W R and Rogers D W O History Overview and recent improvements of EGS4 Report NRC-PIRS-0436

[23] Kawrakow I, Fippel M and Friedrich K 1996 3D electron dose calculation using a Voxel based Monte Carlo algorithm (VMC) Med. Phys. 23 445-457

[24] 2000 VMC++, a MC algorithm optimized for electron and photonbeam dose calculations for RTP Proc. of the 22nd Annual EMBS Int. Conf. Chicago Jul 23-28

[25] Kawrakow I 2000 VMC++, electron and photon Monte Carlo calculations optimized for Radiation Treatment Planning Advanced Monte Carlo for Radiation Physics Particle Transport Simulation and Applications Proc. of the Monte Carlo Conf. Lisbon Oct 23-26

[26] Kawrakow I and Fippel M 2000 Investigation of variance reduction techniques for Monte Carlo photon dose calculation using XVMC Phys. Med. Biol. 45 2163 – 2184

[27] Fix M K, Manser P, Frei D, Volken W, Mini R and Born E J 2007 Efficient photon treatment planning by the use of Swiss Monte Carlo Plan accepted by Journal of Physics: Conference Series

[28] Low D A, Harms W B, Mutic S and Purdy J A 1998 technique for the quantitative evaluation of dose distributions Med. Phys. 25 656-661