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ARTICLE

Transmission portal in vivo dosimetry by means of the Monte Carlo method and the mathematical programming language MATLAB

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Abstract – Modern radiotherapy has increased demand for dose delivery verification. In this paper transmission portal dosimetry was considered. Portal detectors are a promising tool for 2D dosimetric verification and they are nowadays one of the most widely investigated topics. In this study an Electronic Portal Imaging Device (EPID) was positioned below the patient and the transmission images were captured during the irradiation. The principle of this verification consists of comparison of the acquired images with images predicted on the basis of the entrance fluence map and the tissue distribution in the patient. Such verification is not performed at any radiotherapy department in the Czech Republic. There is no system available for the prediction of transmission portal images. Even worldwide, there is still a lack of commercially available solutions. The aim of this paper is to present a new method of prediction of transmission portal images by means of the Monte Carlo (MC) method and the mathematical programming language MATLAB. The MC code EGSnrc (Electron Gamma Shower) was used. The validity of the presented method was verified by comparison of the predicted images with the acquired ones. The acquisition of EPID images was performed at the Hospital Na Bulovce. Three different validation tests were performed. In the first case, the EPID was irradiated by regular and irregular fields while there was nothing present in the beam path. In the second case, a water-equivalent phantom was added to the EPID and was irradiated by a number of irregular fields. In the third case, a real patient was present in the beam path and the EPID images were acquired during the patient's treatment. The patient was irradiated by 8 treatment fields and the portal images were acquired during 5 treatment fractions. All of the acquired images were compared with the MC predicted ones by gamma analysis with gamma criteria of 3%, 3 mm. The average gamma values were 0.31−0.4, 0.34−0.4 and 0.35−0.61 in the first, second and third case, respectively. The results validate the developed method and demonstrate that MC is an effective tool for portal dose image prediction. MC may be favourably used for in vivo transmission dosimetry.

Keywords: patient / dosimetry / scintillation detector / Monte Carlo / Matlab

1 Introduction

Modern radiotherapy techniques achieve high dose distributions to the target volume with steep dose gradients to spare the surrounding healthy tissue. Conformal dose distributions allow treatment dose escalation with the effect of increasing the probability of tumor control. Advances in radiotherapy have increased demand for the quality assurance programmes to ensure that the treatments are delivered correctly (Cufflin *et al.*, 2010). Dose delivery verification is an important issue since correct dose delivery is the basic requirement of successful treatment. Various dosimetric approaches exist, and the dose can be verified either before or during patient irradiation.

In the case of the pre-treatment verifications, errors in the dose calculation of the Treatment Planning System (TPS) can be detected, and correct data transfer can be checked as well as the mechanical and dosimetrical functioning of the linear accelerator. Such verifications concern, for example, the correct realisation of the fluence map and the appropriate functioning of the Multileaf Collimator (MLC). Nevertheless, the pre-treatment measurements do not guarantee accurate fluence delivery during each treatment fraction (van Zijtveld *et al.*, 2009). Moreover, changes in the patient's positioning and anatomy changes cannot be detected. All of these can be checked in the

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dose verifications during treatment (Chin, 2008) and this paper will be dedicated entirely to such verifications.

Currently, many departments verify the dose mainly on the patient's skin at the beam entrance location. The dose is measured by point detectors, mainly by semiconductor diodes or thermoluminescent detectors (Pham, 2009). However, point in vivo dosimetry does not always ensure correct results. The response of the detector is strongly dependent upon precise positioning on the defined place on the patient's skin. The measurement is complicated mainly in regions of high dose gradients (Reich, 2008). Although, some studies exist describing the suitability and accuracy of TLDs for dose determination even in intensity-modulated beams. For example, Engstrom *et al.* (2005) presented an *in vivo* dose verification procedure for IMRT treatments of head and neck cancers. Dose verification was performed by insertion of a flexible naso-oesophageal tube containing TLD rods and markers for EPID and simulator image detection. The measured responses agreed well with the TPS calculations and MC simulations. The use of point detectors may be suitable for treatments of cancers situated in non-moving locations. Nevertheless in moving regions such dosimetry is difficult to use with sufficient accuracy and other methods should be preferred.

A promising approach to dosimetric verification is a 2D dosimetry in which the detector is placed below the patient (Cufflin *et al.*, 2007). Such a method provides more information and is more reliable. Apart from the acquisition of the 2D digital data, the detector has high resolution, short measurement time and high measurement accuracy (Zijtveld *et al.*, 2009). A promising tool for 2D dosimetric verification, applicable even to intensity-modulated beams, is the electronic portal imaging device. The EPID will be considered in this paper. The detector was designed primarily for the alignment of the patient position but nowadays can also serve for quality control of the linear accelerator and for dosimetric purposes (Spezi and Lewis, 2002). The use of the EPID in dosimetry is one of the most widely investigated topics in the medical physics community and a number of studies have been written about this detector. For example, van Elmpt *et al.* (2008) wrote an extensive literature review of electronic portal imaging for radiotherapy dosimetry, summarising information provided in a number of publications over the past 15 years. They discussed dosimetric characteristics and calibration procedures of various types of EPIDs, strategies for using EPIDs for dose verification, clinical approaches to EPID dosimetry and current clinical experience. Despite the favourable detector characteristics and the amount of publications on the subject, there is still a lack of commercially available solutions for EPID dosimetry (Varian Medical Systems, 2007). Several clinics have therefore introduced their own in-house solutions. In Czech hospitals portal in vivo dosimetry is not yet performed since there is no system available for the prediction of transmission portal dose maps. Not even a single study has been published about transmission portal dosimetry in the Czech Republic. This paper should be the first one. The paper intends to present a new method of prediction of transmission portal images by means of the Monte Carlo method and the mathematical programming language MATLAB. The presented work can serve as an inspiration for hospitals equipped with EPID.

There are two main approaches to transmission portal dosimetry. In both, the EPID is placed behind the patient and transmission images are captured during the real treatment.

In the first approach, the transmission portal dose image is predicted and compared with the acquired one (Chin, 2008). This method is called direct transmission dosimetry and will be considered in this paper.

In the second approach, in the inverse method, the measured transmission portal dose image is used to reconstruct the 3D dose delivered in the patient. The reconstructed dose distribution is compared with the distribution predicted by the treatment planning system. The acquired portal image is composed of primary fluence and fluence from scatter. The fluence from scatter is subtracted from the measured transmission map and the transmitted primary fluence then back-projected through the planning CT dataset, yielding the primary fluence within the patient. By convolving this distribution with energy deposition kernels the dose in the patient is calculated. Different approaches exist for the removal of the scatter component, for the back-projection through the CT dataset and for the convolution/superposition. Some of these are discussed in the paper of van Elmpt *et al.* (2008).

2 Materials and methods

This work was carried out on a 6 MV linear accelerator VARIAN® 2100C/D equipped with a scintillation detector EPID aS500.

2.1 Monte Carlo code

Portal images were predicted by the Monte Carlo method, providing the highest accuracy of computation. The system EGSnrc was chosen. The system is a package for the MC simulation of coupled electron-photon transport. Its current energy range of applicability is considered to be 1 keV to 10 GeV (Kawrakow *et al.*, 2011). Extensions of the EGSnrc system are BEAMnrc and DOSXZYnrc.

BEAMnrc is an EGSnrc-based MC simulation code for modelling radiotherapy sources. The system can model all types of linear accelerators using the component modules. One of the outputs of the simulation is a phase space file. This binary file contains data for each particle entering the scoring plane. This is information about the position, direction, charge, energy and weight of each particle (Rogers *et al.*, 2009).

DOSXYZnrc is an EGSnrc-based MC simulation code for calculating dose distributions in a rectilinear voxel phantom. The density and material in every voxel may vary. A variety of beams may be incident on the phantom, including phase space files from BEAMnrc.

The companion program CTcreate is used to create a voxel phantom of a patient from a CT data set of Hounsfield numbers. The generated phantom can be read into DOSXYZnrc and the transport in the phantom can be simulated.

Another program of the EGSnrc package is PEGS4. Its purpose is to generate material data for the EGS code, such as the cross sections of interactions.

2.2 LINAC model

Several studies have emphasised the importance and uniqueness of the use of the Monte Carlo method for the simulation of linear accelerators (Chin *et al.*, 2005; Podgorsak, 2005; Pham, 2009). The MC head simulation results in a precise and accurate description of a radiotherapy beam. The manufacturers of the linear accelerators also use MC codes to optimise their product design (Pham, 2009) and the commercial treatment planning systems use the outputs of MC simulations in their algorithms (Varian Medical Systems, 2008). In this study, the head of the linear accelerator was modelled in BEAMnrc. VARIAN®, the accelerator's manufacturer, provided detailed informations about the dimensions, positions and material compositions of all of the components present in the accelerator head, thus the head was modelled fully without approximations. The electron beam impinging the target was modelled as monoenergetic with Gaussian spatial distribution around the central axis. The necessary parameters of energy and FWHM were determined according to a strategy described in the literature (Sheikh-Bagheri and Rogers, 2002; Pham, 2009). A series of dose profiles and depth dose curves were measured in the water phantom by PTW diode detectors PFD and RFD. SSD was set to 100 cm. The profiles and the depth dose curves were measured for field sizes of $3 \times 3, 6 \times 6$ and 10×10 cm^2 and the profiles were measured in the build-up region of 1.5 cm. The measured curves were compared with the MC-simulated ones. During the modelling different energies (5.7, 6, 6.3, 6.4, and 6.5 MeV) and different FWHM (0.1, 0.15, and 0.2 cm) were considered. The best combination of parameters was then determined according to the analysis of deviations between the measured and simulated curves. The water phantom was modelled in DOSXYZnrc.

2.3 EPID model

The electronic portal imaging device was modelled in DOSXYZnrc (Spezi and Lewis, 2002; Siebers *et al.*, 2004). The scintillation detector EPID consists of 512×384 pixels with space resolution of 0.784 mm, resulting in a detection area of 40×30 cm^2. The cassette of the detector consists of a scintillation layer of gadolinium oxysulphide phosphor and an amorphous silicon flat panel light sensor. Electrons and photons interact in the phosphor layer and form visible light. The light is detected and converted to an electrical signal by a 2D matrix of transistors and photodiodes in the a-Si glass substrate. The amount of the detected charge is proportional to the amount of light striking the diode. For the EPID aS500 it is predicted that 99.5% of the signal comes from the interactions of photons and electrons within the phosphor layer. Therefore, to obtain the response of the EPID from the MC simulation, the energy deposited in the phosphor layer was scored (Spezi and Lewis, 2002; Siebers *et al.*, 2004).

The EPID was modelled according to informations provided by the producer. These included geometry and the chemical composition of individual layers. The library of cross sections in EGSnrc did not contain data for most materials present in the detector (out of cover, air, nickel, copper, FR4, rohacell, foam, paper, poly, screen and glass only air, nickel and copper were defined in the library). A new file was created in the PEGS4 program containing cross section data for all materials present in the EPID (Rogers *et al.*, 2009).

The EPID model was verified by comparing measured and simulated responses in the detection layer of the detector (SDD 105 cm) for 4 regular ($3 \times 3, 5 \times 5, 10 \times 10$ and 15×15 cm^2) and 11 irregular fields shaped by MLC. Gamma analysis was considered according to Low *et al.* (1998).

Prior to the EPID acquisitions the detector was calibrated dosimetrically according to procedures described in the VARIAN manual (Varian Medical Systems, 2007). Firstly, a dark field image was captured to determine the response of every pixel in the matrix on background. Secondly, a flood field image was captured. The whole sensitive area was irradiated to determine the relative sensitivity of every pixel in the matrix. Thirdly, correction was made to preserve the dosimetric properties of the beam. The portal image was multiplied by a radial symmetrical correction matrix generated from a diagonal profile of the largest field size. And lastly, the portal imager was calibrated so that the signals generated by the radiation beam were related to the dose measured by the reference ionisation chamber. The acquired portal doses were then displayed in calibration units (CU).

2.4 Patient model

A phantom of the patient was created in the program CT create from the patient's CT slices. To each voxel of the phantom, a value of density and material was assigned from the knowledge of its CT number. The default conversion curve of the CTcreate could be used (Rogers *et al.*, 2009) however, since the curve is sensitive to the Roentgen spectra, it was more appropriate to acquire our own conversion curve characteristic of the given CT scanner. In this study, the CT was calibrated on the basis of tissue-equivalent samples and stoichiometric calibration (Schneider *et al.*, 1996). The phantom of the patient was created out of 4 materials (air, lung, tissue and bone) with a grid size of $0.2 \times 0.2 \times 0.2$ cm^3. The grid size was a compromise between computation time and resolution. The created phantom was validated against the CT slices visually and by size comparison of various organs.

Afterwards, dose distributions were predicted inside the patient for various treatment fields. These MC predictions were then compared with the predictions from the planning system. The analysis of agreement was performed on the basis of visual comparison of isodoses (Chow and Leung, 2006, 2008).

2.5 Transmission through a RW3 phantom

Transmission through a water-equivalent RW3 phantom was investigated. The RW3 solid slab phantom ($30 \times 30 \times 5$ cm^3) was placed on the EPID and was irradiated by 8 irregular fields with the gantry at 0°. The portal dose maps were acquired at SDD of 105 cm and were compared with the MC predicted ones by gamma analysis.

2.6 Transmission through a real patient

The prediction of transmission portal images requires a patient and the EPID to be modelled in the beam path. This is problematic. The phantom of the patient is created in the program CTcreate, while the EPID is modelled in DOSXYZnrc. The DOSXYZnrc enables only either the patient or the EPID to be placed in the beam. The limitation was solved by creation of a combined phantom which is composed of both the patient and the EPID. For this purpose a script in MATLAB was written.

Once the combined phantom was created, the possibility of irradiating the patient from any arbitrary gantry angle was considered. The situation in which the patient lies on the treatment couch, while the gantry and the EPID are rotated by angle theta, was simplified. Geometrically, the irradiation of the patient from angle theta is the same as the irradiation of the patient from 0°, in the case that the patient is rotated by negative angle theta. The EPID is fixed with the accelerator head, therefore not only the head but also the EPID stays without any rotation (Chin *et al.*, 2003; Kairn *et al.*, 2009).

Scripts in MATLAB were written to rotate the patient phantom by any angle, to add the EPID beneath the rotated phantom and to cut the combined phantom according to the region of interest.

The last step was focused on direct transmission dosimetry with a real patient in the beam path. The EPID was placed behind the patient at SDD of 140 cm. The patient was irradiated by 8 treatment fields per fraction and the portal dose images were acquired during 5 treatment fractions. Under the same conditions, transmission portal images were predicted by the new developed method. The acquired and the predicted images were compared and evaluated by gamma analysis with gamma criteria of 3%, 3 mm.

3 Results

3.1 LINAC model

The head of the linear accelerator VARIAN® 2100C/D was modelled in BEAMnrc (Fig. 1). A photon beam of nominal energy of 6 MV was considered. In the figure different components, their material composition and their geometrical positions are indicated. No approximations to the model were done as the manufacturer provided detailed geometric data and the material specification for all of the head components and accessories.

The best combination of parameters (E and FWHM) for the electron beam impinging the target was carefully selected as these parameters have a significant effect on the generated Bremsstrahlung photon beam exiting the target. This effect was studied in detail by Sheikh-Bagheri and Rogers (2002) using the BEAMnrc and DOSXYZnrc Monte Carlo codes. In this study, a similar procedure was used to find the two critical parameters. The electron energy and FWHM were tuned by matching the simulated and measured PDDs and dose profiles. It was found that the FWHM of the electron beam did not affect the PDD at all. The PDD was only affected by the initial electron beam energy. The higher the energy, the greater the

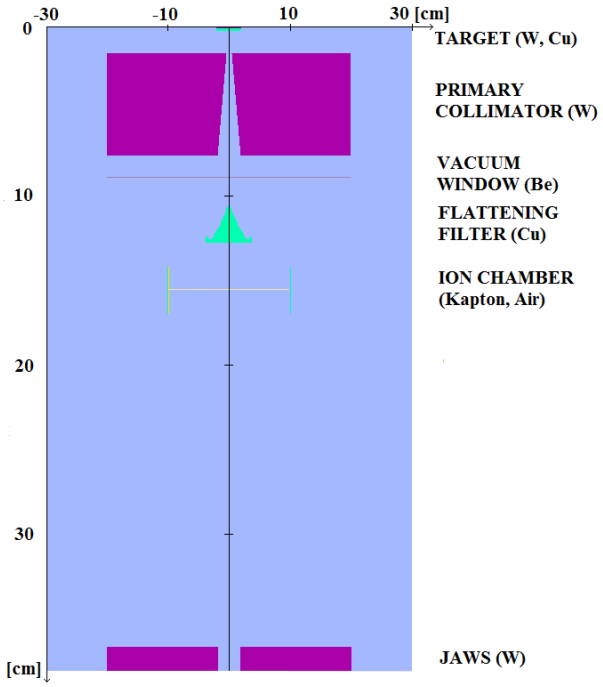

Fig. 1. Model of the linear accelerator head.

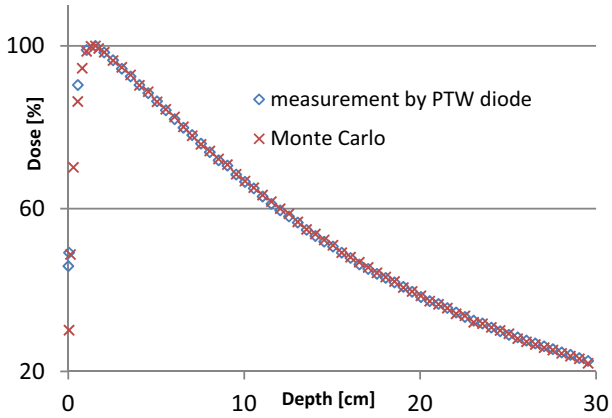

Fig. 2. PDD curves of the tuned beam (E 6.3 MeV, FWHM 0.1 cm, field 10×10 cm^2). Size of error bars is smaller than the dimension of the plotted marks.

depth of dose maximum. The dose profiles on the other hand were strongly affected by the FWHM. The smaller the FWHM value, the sharper the penumbra. E of 6.3 MeV and FWHM of 0.1 cm gave the closest agreement between the modelled and measured dose data and these were used for all successive simulations. The agreement for the tuned beam can be seen in graphs of PDDs and dose profiles (Figs. 2 and 3).

3.2 EPID model

The electronic portal imaging device was modelled with pixel size of individual layers of 0.2×0.2 cm^2. The model was verified by measurement of responses to 4 regular and 11 irregular static fields. Prior to the image acquisition the EPID was calibrated dosimetrically according to the procedure described

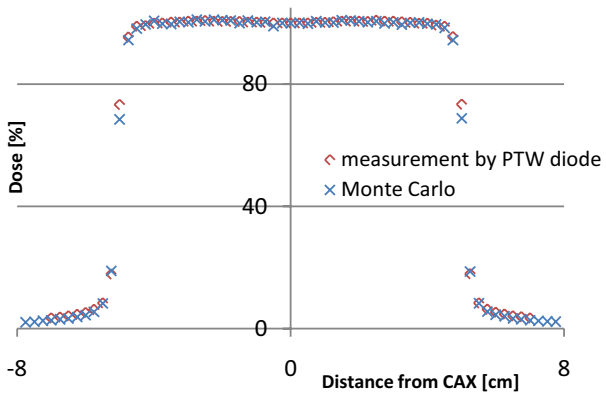

Fig. 3. Profiles of the tuned beam (E 6.3 MeV, FWHM 0.1 cm, field 10×10 cm^2). Size of error bars is smaller than the dimension of the plotted marks.

Table 1. Average gamma in relation to the material present above the EPID.

Material present in the beam path	$\bar{\gamma}$ (3% ΔD, 3 mm Δd)
–	0.31–0.4
RW3 phantom	0.34–0.4
Patient	0.35–0.61

in Section 2.3. The agreement between the measured and computed images was satisfactory. For the criteria of 3%, 3 mm the average gamma values were in between 0.31 and 0.4 (Tab. 1). During the gamma analysis a region of interest was selected including the whole treatment field. Siebers *et al.* (2004) reported a similar study in which a MC EPID model was tested for a 6 MV photon beam for various field sizes (5×5, 7×7, 10×10, 15×15 and 20×20 cm^2). SDD was set to 105 cm. The predicted portal images were compared with EPID measurements performed under the same conditions. Profiles were compared along the x and y directions through the measured and computed images. For each field size, the profile along the left-right direction was accurately predicted within less than 1% for all pixels. The small deviations observed were attributed to the statistical uncertainty of the MC simulation. In the anteroposterior direction similar results were obtained for the fields up to 15×15 cm^2. For larger field sizes, the measured and computed profiles did not agree near the field edge on the gantry side of the profile up to 3%. The region of disagreement corresponds with the location of the R-arm mounting bracket and the associated hardware beneath the imager. These materials provide nonuniform backscatter to the imaging cassete during the EPID calibration. The calculations utilised a uniform backscattering material slab and did not model the R-arm mounting bracket. In this work, to account for the backscattering, a uniform material slab was also modelled. However, the described effect was not observed as field sizes larger than 15 cm were not tested. Siebers *et al.* (2004) also described deviation outside the field but it was not considered significant since it corresponded with less than 0.15% of the field dose. The possible causes of this discrepancy are uncertainties in the portal imager background subtraction, ghosting on the portal imager or calculation errors. In this paper only the

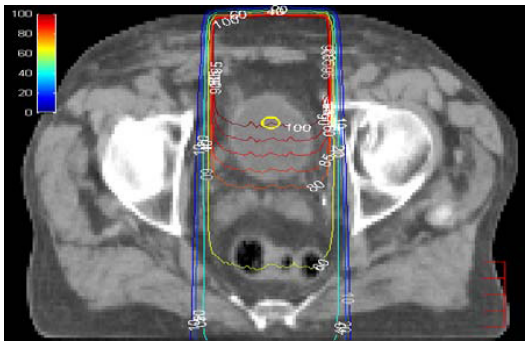

Fig. 4. Dose distribution predicted by MC method.

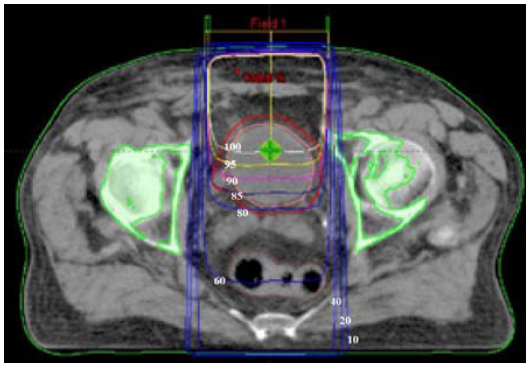

Fig. 5. Dose distribution predicted by treatment planning system Eclipse.

treatment field was taken into account during the gamma analysis Deviations outside the field were therefore not detected.

3.3 Patient model

A phantom of the patient with voxel size of $0.2 \times 0.2 \times 0.2$ cm^3 was created (Fig. 4) from the patient's CT slices (Fig. 5). The phantom was validated against the CT slices visually and by measurements. Sizes of the real and modelled organs were measured and agreed well within the phantom's voxel size.

Subsequently, dose distributions were simulated in the patient for different treatment fields. The achieved statistical uncertainty with a 2σ significance level was lower than 2% isodose lines of these distributions (Fig. 4) were compared with isodose lines of distributions obtained from the TPS Eclipse (Fig. 5). In the TPS a calculation grid of 2 mm was selected. Isodose lines from MC and TPS predictions were plotted separately in the two figures. The reason is that dose distributions predicted by MC were imported on top of the voxel phantom, while dose distributions predicted by TPS were imported on top of the CT slices. However, from the visual comparison of isodoses a satisfactory agreement can be observed.

3.4 Transmission through a RW3 phantom

Prior to testing transmission through the patient, transmission through a homogeneous water-equivalent phantom

($30 \times 30 \times 5$ cm^3) was considered. The aim of this procedure was to verify the presented method when additional material has been inserted in the beam path. The phantom was placed on the EPID at SDD of 105 cm and was irradiated by 8 irregular fields. The RW3 phantom was modelled with voxel size of $0.2 \times 0.2 \times 0.2$ cm^3. The measured and predicted portal images were compared by gamma analysis. For the criteria of 3%, 3 mm the obtained average gamma values were in between 0.34 and 0.4 (Tab. 1). The γ results were very similar and only slightly worse when compared with the γ results obtained in Section 3.2 ($\bar{\gamma} \epsilon \{0.31, 0.4\}$). The observed difference was in agreement with expectation since in this case the RW3 phantom is causing additional attenuation and scatter.

3.5 Transmission through a real patient

The developed transmission dosimetry method was tested on a real patient with 8 treatment fields per fraction. The transmission portal images were acquired during 5 treatment fractions. The images were compared with the predicted ones by gamma analysis. The results were satisfactory; for the criteria of 3%, 3 mm, the average gamma values were in between 0.35 and 0.61 (Tab. 1). The largest discrepancies between measured and predicted images occurred at field edges and at other regions of high dose gradients. Between fractions the average gamma values differed mostly by 0.2. Some of the discrepancies observed between fractions can be attributed to changes in patient geometry, changes in patient setup and changes in the output of the linear accelerator. On the other hand, the stochastic nature of the MC simulation and the simplified patient model influence the precision and accuracy of the predicted portal images.

In the case of transmission dosimetry, tolerances on the average gamma values, the criteria for the quality evaluation, were not determined. For this purpose it would be necessary to gather and evaluate maps for an extensive set of patients and sufficient number of fractions. This could be a theme for future work.

4 Conclusion

A method enabling the verification of correct patient irradiation was developed and verified. Three different validation tests were performed. In these tests the acquired EPID images were compared by gamma analysis with the images predicted by the MC method. Firstly, no material was present in the beam path; secondly, a water-equivalent phantom was placed on the EPID and lastly, transmission dosimetry was tested on a real patient. For the 3 cases, for the gamma criteria of 3%, 3 mm, the average gamma values obtained were 0.31−0.4, 0.34−0.4 and 0.35−0.61, respectively. The achieved statistical uncertainty of the MC calculation with a significance level of two sigma was lower than 2%. The results confirmed that a full MC simulation of particle transport will correctly predict the acquired EPID images. In the paper of Siebers *et al.* (2004), in which the MC EPID model was tested without the patient in the beam, similar conclusions were reported. It was concluded that MC dose calculation can accurately predict portal dose maps.

The presented method enables verification of correct patient irradiation for every treatment field in the real time of treatment. Transmission dosimetry is very useful. Apart from the verification of correct realisation of the fluence map, the functionality of the accelerator can be checked from the mechanical and dosimetrical point of view. The correct and reproducible localisation of the target volume and the critical structures can also be checked. However, worldwide, there is a limited number of commercial systems for portal dosimetry (Varian Medical Systems, 2007) and many hospitals have developed their own solutions. In the Czech Republic the transmission dosimetry is not performed at any department in the country since there is no system available for the prediction of transmission portal images.

The presented methodology is beneficial and can be used in clinical practice to increase the quality of radiotherapy. However with respect to the high time demands of MC simulation, the method can be used only for a limited number of patients. Children would be one of the target groups for which such verification could be systematically used since the tissues and organs in children are growing fast and are more sensitive to irradiation. The tasks in the presented method were solved in parallel by a multi-core processor. Nevertheless in future attention should be paid to cluster calculation.

Other suggestion for future work would be the evaluation of results of transmission verification for a number of patients. Besides the results of transmission verification, the success of the treatment would be monitored. The patients would be divided into groups according to their diagnosis, spread of disease (classification TNM) and maybe according to their age and gender. On the basis of the evaluation of a large data set it would be possible to determine tolerances for the parameter of the average gamma value. The tolerances obtained would be used for future qualitative analysis of patient irradiation.

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