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WP13 – M2 First 3D dose computation

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1. Introduction

Calculation of the dose in a single point under well defined conditions can be perform easily with the use of calculator, however calculating the distribution of dose in a patient is not straightforward. Due to the complexity of radiation interaction with the human tissues and due to the practical need for rapid calculation times, such dose calculation is a complicated process, which has also its limitations due to the approximations used in the physical models. For those reasons dose calculation algorithms are the most critical software components in a computerized treatment planning software (TPS). Their ultimate goal is to predict accurately the dose delivered at any single point in the patient in order to decide if a given treatment plan is acceptable or to choose the best plan from several alternative ones [1], [2].

1.1 Historical perspective

Before computerized treatment planning systems became widely available, dose distributions were calculated manually by the addition of percentages estimated from the superposition of isodose charts (such as those shown in Figure 1.1). The computation methods involved empirically derived corrections to account for patient shape and inhomogeneities. It was a long and time consuming process, which depended on the planners experience [1].

![Figure 1.1 Isodose lines in the central plane (left) from a cobalt-60 beam and (right) from a 6 MV linear accelerator][1]

First computers with software dedicated for radiotherapy (RT) were in-house developed for individual purposes. A pioneer, where an automatic computing machine for radiation dosage calculation was used for a first time (1955), is a Memorial Hospital in New York. 1060 was the year, when a digital computer and a method of dose calculation, that was originally developed for manual calculations by Clarkson, was applied in RT [3]. The first dose calculation algorithms for the treatment planning of proton radiotherapy were written for scattered beam techniques and use a broad-beam technique proposed by Chen in 1979 [4]. Evolution in computer technology led not only to development of dose algorithms but also diagnostic imaging. The simultaneous progress in computed tomography (CT) and increasing computing power in 1970s was the basis for computerized treatment planning, as it is understood today. CT scanning was ideally suited for radiation therapy planning, since it provided for the first time, an easy ability to localize the tumor and surrounding normal tissues on the patient’s axial anatomy. Moreover, it provided ability for calculating dose distributions, which account for the real tissue densities within the patient [2], [5]. Modern treatment planning has evolved and full 3D planning techniques are available including patient data from a variety of different imaging sources (CT, MR, US, SPECT, PET). Commercial systems based on the published algorithms are now widely available and as the power of computer technology grows, more complicated dose algorithms are generally in use.

1.2 General requirements

New dose delivery techniques in radiotherapy, such as the development of intensity-modulated radiotherapy (IMRT) or the implementation of spot scanning for ion beams,
have enhanced potential of conforming 3D dose distribution to identified targets while organ at risk (OAR) can be spared. Those techniques, however, require fast and accurate dose algorithms, which can be applied for treatment plan optimization in clinically acceptable timescales. The crucial role of the dose calculation is seen especially when the considered clinical case contains complex tissue inhomogeneities. Heterogeneities in particle therapy are much more important than in photon therapy. As the range depend on the heterogeneities, than the actual delivered dose will be sensitive to the exact positioning of those heterogeneities in relation to the particle beam [6], [7]. The complicated patient anatomy requires very fine and accurate sampling of particle interactions with the considered tissues, which result in time-consuming calculation. Not only approximations in the modeling of radiation interactions or uncertainty in the basic radiation interaction data has an influence on the calculation of the dose distribution within the patient. Precise calculations depend also on proper combination of the relevant beam data with individualized patient data and the calculation grid size, which is the interval between beam calculation points, and also accurate interpolation from the beam calculation grid to the patient anatomy grid [8]. Dose calculation, however, is only one of the steps in the whole radiotherapy process. Table 1.1 illustrates the uncertainties attached to each of the elements which contributes to the final accuracy in radiotherapy.

Table 1.1 Contributions to the accuracy in radiotherapy [9]

<table>
<thead>
<tr>
<th>Determination of the Accuracy Required for Dose Calculation</th>
<th>Present Technique</th>
<th>Future Development</th>
</tr>
</thead>
<tbody>
<tr>
<td>Absorbed dose determination at the calibration point</td>
<td>2.0</td>
<td>1.0</td>
</tr>
<tr>
<td>Additional uncertainty for other points</td>
<td>1.1</td>
<td>0.5</td>
</tr>
<tr>
<td>Monitor stability</td>
<td>1.0</td>
<td>0.5</td>
</tr>
<tr>
<td>Beam Flatness</td>
<td>1.5</td>
<td>0.8</td>
</tr>
<tr>
<td>Patient data uncertainty</td>
<td>1.5</td>
<td>1.0</td>
</tr>
<tr>
<td>Beam and patient set-up</td>
<td>2.5</td>
<td>1.6</td>
</tr>
<tr>
<td>Overall excluding dose calculation</td>
<td>4.1</td>
<td>2.4</td>
</tr>
<tr>
<td>Dose calculation</td>
<td>1.0 2.0 3.0 4.0 5.0</td>
<td>0.5 1.0 2.0 3.0 4.0</td>
</tr>
<tr>
<td>Resulting overall uncertainty</td>
<td>4.2 4.6 5.1 5.7 6.5</td>
<td>2.4 2.6 3.1 3.8 4.7</td>
</tr>
</tbody>
</table>

Uncertainties are expressed as percentages. The present technique refers to 1999.

Two situations are considered the present technique (in 1999) and the future development. Several hypothetical levels of dose computation uncertainty (from 0.5% to 5%) are considered. And it turn out that a dose calculation uncertainty of about 2% or 3% does not lead to a significant increase in the overall uncertainty at present, whereas 1% calculation accuracy might be significant in the future [1].

## 2. Algorithm classification

Over the years, dose calculation algorithms evolved from one-dimensional to fully 3D techniques. However, it is important to distinguish the difference between 3D display of dose distributions and 3D calculation of the dose. The real 3D calculation is when the primary and scattered radiation components are followed independently throughout the volume of irradiated tissue, it accounts for any change in the whole volume, while 3D display in the TPS gives the illusion of the user moving through 3D object in 2D representation [5].

There is no clear method of classification for dose calculation algorithms. ICRU (1987) distinguished between tabular (or matrix) formats, beam generating functions, separation of primary and scattered radiation and representations using basic principles. Others propose to distinguish between correction-based and model-based methods. However, the easiest categorization, which comes to mind, is a suggestion to classify algorithms according to the number of dimensions over which the integration is carried out: the absence of integration was referred to as the broad-beam approach, because the beam is treated as a whole rather than being broken up into its component segments (Figure 1.2). The Monte-Carlo approach, where each particle is tracked in 3D, does not fit exactly with this scheme. It could be considered as the ultimate form of the 3D integration and therefore treated separately.
This approach provides a classification into 1D, 2D, 2.5D or 3D models. Typically, for a 1D algorithm the dose at point P depends only on the accessories and patient characteristics found on the line joining source to P; 2D algorithm takes into account the modifications in the transverse plane, excluding the changes in the caudo-cranial direction, whereas 3D accounts for any change in the whole volume [1].

2.1 Broad beam method

The dose representation of simple beams (i.e. rectangular fields limited by the main collimator without any accessories) in water can be further subdivided into two types:

- Representations that use tabulated beam data dose distributions for a number of beams measured under reference conditions are stored in tabular form and are interpolated by the treatment planning system (TPS) during calculation.
- Representations that use an analytical approach (beam generating functions): depth-dose and profile characteristics are modelled using mathematical functions. Some of these approaches are purely mathematical, while others more closely reflect the physics of broad beam interactions.

Typically, for particle therapy, these may consist of depth-dose curves for spread-out Bragg peak (SOBP) together with lateral dose profiles at representative depths. From these data, the dose at any point in the distribution can be derived by calculating its water equivalent depth (the integral in depth of the water equivalent densities, from the patient surface down to the point of interest), usually performed using a ray tracing algorithm, and interpolating the resulting dose from the measured depth-dose data. This method can be used for both scattered and wobbled beams (Figure 2.2). Broad beam method is the simplest, fastest, and least precise approach for estimating dose. It requires applying many corrections to the original measured data in order for it to be accurate in real situations involving patients [1], [11]. Today this algorithm is used only for ocular applications with protons.

2.2 Pencil beam method

In order to use the advantages of particle therapy, active scanning treatment methods have been developed and pencil beam (PB) algorithms are used here to compute the dose.
Compared to passive scattering techniques, where a small number of broad fields are used, many (up to tens of thousands) narrow particle beams (spots) are used for irradiation in active scanning. With pencil beam algorithms more accurate modelling of dose can be achieved.

Typically, the incident beam is modelled using a number of closely spaced finite pencil beams, with each pencil beam being assigned a weight that is directly proportional to the particle fluence of the beam for the pencil’s position. Each pencil beam broadens because of multiple Coulomb scattering within the patient, and its lateral shape can be modelled using measured or calculated data. The resultant dose at any point is then computed by summing the contributions from each of the pencil beams, with each calculated point taken to be its actual water-equivalent depth [11], [12]. There are many pencil beam algorithms available on the market and basically each vendor has developed its own PB algorithm. In my study I am using the XiO treatment planning system provided by ELEKTA Software, where a proton pencil beam algorithm is implemented. A more precise explanation of the algorithm is presented in chapter 3.2.2.

2.3 Monte Carlo algorithms (MC algorithms)

For radiotherapy with both photons and particles, there is a strong move towards Monte Carlo (MC)-based dose calculation engines, especially when calculating dose distributions for intensity modulated radiotherapy. It is because the calculation based on MC simulations is the most reliable method for calculating the detailed transport of particles in tissues.

In this approach, individual particles are tracked as they penetrate through the patient and interact with the material through which they pass. The likelihood of an interaction, and its consequences, is sampled using random numbers, from the best available probability distributions. For MC algorithms it is easy to model Coulomb interactions that lead to energy loss and scattering of particles. However, to obtain sufficient statistical accuracy for useful dose distributions in practical situations, tens of millions of histories usually must be traced. Such calculations can take hours or even days to process [11], [13]. In near future MC algorithms will be preferred for treatment planning in particle therapy.

Recently, a simplified Monte Carlo (SMC) planning algorithm for proton beams based on measured depth dose distributions in water has been developed. Relatively short calculation time SMC agrees well with experimental results in a heterogeneous phantom. However, further investigations will be necessary to quantify the calculation time as well as the influence of these simplifications on the results. Another option is the implementation of established general purpose MC codes like GEANT4 or FLUKA for treatment planning. However, these codes are still too slow for clinical dose calculations. For particle therapy treatment planning, fast and accurate MC dose calculation algorithms are currently being developed [11,14,15].

3. Treatment planning software - description

As a tool to create treatment plans for protons with the spot scanning technique, I use the XiO treatment planning software provided by ELEKTA Software.

3.1 XiO- general information

XiO is a family of 3D and IMRT radiation treatment planning systems used to develop plans for cancer patients. Apart of proton planning, XiO supports a range of other treatment modalities, including 2D, 3D, MLC-based IMRT, solid compensator-based IMRT and brachytherapy. Dynamic conformal arc therapy and stereotactic delivery also are supported. With automated image fusion, XiO delivers rapid and reliable registration of multiple data sets. Automated contouring tools, including patented auto-segmentation functionality and powerful drawing and editing features, allow quick and easy identification and delineation of target volumes and critical structures. The dynamic 3D visualization and surface rendering allows the display of patient anatomy and beam parameters in real-time. XiO uses DICOM services to import images, structures and plan parameters and to export
images, structures, plan parameters and dose to other vendors. It supports the network import of CT, MR and PET images, RT Structure Sets and RT Plans. The dose calculation is flexible by choosing the appropriate calculation algorithm out of multiple 2D and 3D algorithms. For External Beam Application Software for Photon and Electron Treatment Planning user may choose from:

- Clarkson Algorithm
- FFT Convolution Algorithm
- Superposition Algorithm
- 3-D Pencil Beam Algorithm
- Irregular Field Point Dose Calculations

Within proton, user may use the following:

- Broad Beam supports uniform scanning, wobbled beams and scattered beams. Spread-out Bragg Peak (SOBP) can be created via rotating wheels and ripple filters. Apertures and range compensators are supported. Here the broad beam or pencil beam algorithm for dose calculation is available.
- Spot Scanning with Intensity Modulated Proton Therapy (IMPT) includes nuclear interactions, subspots, apertures and range shifters [16], where the proton pencil beam algorithm is used. This is the method and dose algorithm I use to perform my study.

Other features provided by the vendor are:

- DRR Viewing and Output
- Dose Volume Histograms (DVH)
- TCP and NTCP Calculations
- Fixed, Motorized, Dynamic and Virtual Wedges
- Siemens lmFAST MLC segmentation software.

### 3.2 XiO- pencil beam for protons

Proton dose calculations in XiO are based on the method described in the paper "A pencil beam algorithm for proton dose calculations" by Linda Hong, Michael Goitein, Phys. Med. Biol. 41 (1996), p. 1305-1330, [17], which is similar to the Gaussian pencil beam model used for electron beams formulated by Hogstrom et al (1981) [18].

#### 3.2.1 Physical properties modelled

Protons in the energy range of interest for radiation therapy (when they passing through the all beam modifying devices and the through the patient) undergo two main processes: energy loss and small-angle scattering. Protons lose energy primarily through multiple Coulomb interactions [19]. The multiple scattering of protons by the material through which they pass causes angular standard deviation of the distribution in the beam which ultimately increases the size of the penumbra and affects the way internal and external inhomogeneities perturb the dose. The shape of the angular distribution as the result of the multiple Coulomb scattering is nearly Gaussian for small angles [17].

![Figure 3.1 Beam modifier locations](image)
Figure 3.1 is a diagram showing the beam modifier locations that are used in the dose calculations. As protons reach each beam-modifying device, its influence on the beam is computed. All distances are to the downstream face of the device. The air gap is the distance along the beam central axis between the downstream-most face of the downstream-most beam modifier (which will be the range compensator or compensating block if one exists) and the patient. The effect on the penumbra due to beam modifiers such as degraders and modulators that are located upstream of the aperture is modeled by:

* Computing the angular confusion introduced by each modifier
* Projecting it to the source location
* Adding its contribution in quadrature to the effective source size

The angular confusion introduced by a degrader or modulator element is different for each element of the modulator and is dependent on these factors:

* Location of the downstream face of the modifier
* The thickness of the modifier
  * The material type (Lucite or water)
  * The input range prior to entry into the modifier

Degraders, modulators, range compensators, and compensating blocks are classified as "intermediate thickness" materials because they do not absorb all the energy of the proton beam. Their angular distribution is therefore taken to be a Gaussian with characteristic scattering angle, which is given by the generalized Highland approximation as described in the paper "Multiple Coulomb scattering of 160 MeV protons" [19]. Beam-limiting devices (collimators, apertures and blocks) are assumed to completely absorb protons, which hit them, and to transmit all other protons without affecting them. Thus those particles, which scatter out of a beam-limiting device before losing all their energy, were ignored [17].

Multiple Coulomb Scattering in the Patient is different from all other elements because one generally wishes to know the dose at arbitrarily chosen points within him or her. The patient is classified as an "infinitely thick" material since all the proton energy is absorbed in the patient. The radial distribution of fluence and dose at depth within the patient has a Gaussian distribution and can also be calculated using the generalized Highland approximation [19].

The final total multiple Coulomb scattering or angular confusion is calculated by adding in quadrature the multiple Coulomb scattering contributions at the dose point.

3.2.2 Pencil Beam Algorithm

The pencil beam algorithm approximates the full beam by a set of elemental pencil beams. The dose at any point is taken to be the sum of the doses delivered by each of the pencil beams. The algorithm is composed then of two parts: the computation of the dose at a point of interest from a given pencil beam; and the technique for summing the contributions of the pencil beams.

To compute the dose at a point of interest from a given pencil beam the Hogstrom's electron pencil beam algorithm is followed:

- The dose distribution is separated into a central-axis and off-axis term.
- The central-axis term is related to the measured Bragg peak by an effective depth and inverse square correction.
- The off-axis distribution is described by a Gaussian distribution whose standard deviation is found by adding in quadrature the multiple Coulomb-scattering contributions from all devices in the beamline [17].

In summing the pencil beams, one must determine how to sample them. For every point of interest for which the dose is to be calculated, a polar grid in a plane perpendicular to the central axis of the incident beam is set up. The first step of the pencil beam loop is to determine the maximum radius around the dose point over which pencil
contributions need to be included. This radius is equal to $3\sigma$, where $\sigma$ is the angular confusion at the dose point for the maximum range compensator thickness and maximum range in patient. The maximum possible angular confusion must be used because it may represent the angular confusion of neighboring pencils [20].

$$\sum_{i=-\infty}^{+\infty} \text{erf}(x_i) - \text{erf}(x_{i-1}) = 1.0$$  \hfill (1)

The pencil beam algorithm is a "receiver mode" calculation, which increases the computation time linearly as a function of the number of points being calculated. Also, the central-axis term (CAX) at each sector surrounding the point of interest is calculated. The contribution from that sector to the dose point equals that sector’s central-axis term times the pencil factor:

$$CAX_j(p) = \text{depth}_\text{dose}_j \times \text{inv}_\text{sqr}_j$$  \hfill (2)

$$\text{dose}(p) = \text{weight}_j \times \sum_{j=1}^{N\_PENCILS} CAX_j(p) \times \text{factor\_pencil}_j(p)$$  \hfill (3)

$$\text{dose}_j(p) = \sum_{l=1}^{N\_PENCILS} \text{dose}_l$$  \hfill (4)

N\_PENCILS -Total number of pencils
j -Individual pencil index
factor\_pencil\_j - Fractional contribution of pencil to dose point [20]

3.2.3 Pencil beam algorithm limitations

**Slit Scattering**
Protons which scatter off the edges of the collimator and aperture can contribute significant dose near the patient surface. This effect is largest for small air gaps and small fields. For larger fields, the effect causes "horns" or "ears" to appear in lateral distributions near the surface.

**Small-Angle Approximation**
The algorithms make a small-angle approximation and ignore wide-angle scattering. This approximation is justified because only a small amount of energy is contained in the tails of the distributions.

**Shadows of Thick Heterogeneities**
Like the electron pencil beam algorithm, the proton pencil beam algorithm produces errors in the shadows of thick heterogeneities [17,20].
3.3 XiO- optimization process

Optimization routines are provided by TPSs with varying degrees of complexity. Algorithms can modify beam weights and geometry or calculate beams with a modulated beam intensity to satisfy the treatment objectives. Plan optimization is a process of iteratively generating and then automatically assessing a large number of plans and choosing the best among them. The mathematical meaning is to choose those values for treatment-delivery variables that would result in an extremum of the score function. Intensity modulated proton therapy (IMPT) is a scanning technique using the approach of inverse treatment planning. Each spot is weighted in a computer-based optimization procedure to find the optimal solution for the prescribed dose in the target volume and for sparing the critical organs as much as possible [14]. In forward planning, on the other hand, the treatment planner needs to optimize beam parameters and evaluate dose distribution more or less in a trial and error process (Figure 3.3).

![Figure 3.3 Schematic interpretation of forward planning (left) and inverse planning (right) [21]](image)

Several optimization algorithms have been proposed for radiation therapy. The one, which has been implemented in XiO tps is precisely described in my previous deliverable about “Basic principles for treatment planning software for light ions” [22].

4. First 3D dose computation

At the beginning of the study, first 3D dose computation was perform using the algorithm described above. To see the variety between dose distributions in the patient, three separate plans were created with different beams incidence.

4.1 Data preparation

For planning purpose CT data set for a patient treated with prostate cancer was selected. The CT was taken in the treatment position with 4mm slice thickness. Organ at risk (bladder and rectum) as well as clinical target volume (CTV) was delineated. CTV was considered as a prostate without seminal vesicles and it was expanded to a planning target volume (PTV) by adding appropriate margins: 5 mm in the axial and 8 mm in the craniocaudal dimensions. A greater margin was used in the superoinferior direction to account for the larger uncertainties in identifying the prostatic apex and base that could affect changes in the PTV coverage inferiorly and superiorly [23]. Figure 4.1 illustrates all delineated structures on patient’s CT. In all 3 projections the bladder is shown in yellow, the rectum in cyan, the CTV in red and the PTV in magenta. Normally, for planning CT

![Figure 4.1 Structures delineated on patient’s CT shown in 3 projections](image)
purpose and in order to see the bladder better, patient gets an intravenous contrast media. To simulate the bladder without the contrast agent, the electron density of the opacified bladder was virtually changed to water density (which is equal 1). The same procedure was performed for the rectum, because as a standard procedure in proton therapy, a rectum balloon filled with water is inserted, to immobilize the prostate.

### 4.2 Treatment planning

Three separate treatment plans were created using the spot scanning technique, with one (90°) beam, 2 opposed lateral beams (90°, 270°) and 3 beams (0°,135°,225°), always centred on the PTV. The spot scanning technique for protons allow us to deliver the dose in spots, starting from the distal edge of the tumour (Figure 4.2b). The use of several beam angles results in excellent, 3-D conformation of the high dose to the target volume. In order to create a beam, different parameters must be defined (all seen in Figure 4.2a). After entering the parameters like peak with multiplier (0.8), alignment shift (0.5cm), spot sigma (0.3cm) and spot spacing (0.6cm) which describe the properties of the spots the TPS algorithm automatically defines additional spots in distal and lateral direction around the PTV, their range and also intensity.

Once beams are created, the dose can be calculated. For each plan the calculation grid size for dose calculation was set for $3 \times 3 \times 3$ mm$^3$. The prescribed dose for every plan was 78 Gy to the PTV. In Figure 4.2 constrains for target, bladder and rectum are presented for dose optimization purposes. All delineated structures are listed there but only 3 are active for the optimization process. PTVct is a target and has the highest rank (1), rectum and bladder have a lower rank (2, 3 accordingly). That means, whenever those structures overlap each other, contribution of voxels in the volume with the highest rank will be included in the objective function. The next step is to define dose constrains which the optimization algorithm will try to reach. With the method of trial and error the best possible constrains are chosen in order to meet the planners expectations. Here the plan was consider as good when 95% of the prescribed dose went to 95% of the PTV and 100% of the CTV. For the target, maximum (7850cGy) and minimum (7750cGy) were set with the weight 100. The weight is an option to increase the relative importance of an objective. For bladder the maximum allowed dose was 7800cGy, also 10% of bladder volume should not receive more than 5000cGy and 20% of volume should not receive more than 2000cGy (weight 300). For rectum the maximum dose constraint is also 7800cGy, 7% of volume should have below 5000cGy and 17% should have less than 2000cGy. Sparing rectum is more important than bladder, that is why the weight for the rectum (400) was set a little higher than for the bladder (300). Finally, the dose algorithm is ready to calculate the dose.
and then the optimization algorithm will optimize the weight of each spot in order to satisfy the treatment objectives.

4.3 Calculated dose distribution

After calculation of the dose the isodose distribution is verified to ensure that target coverage is adequate and that critical structures surrounding the PTV are spared as necessary. The XiO system allows the use to scroll through the slices in all 3 projections and also display the dose in the beam eye view (BEV) and 3D representation. Finally, it allows to create dose volume histograms (DVH) (Figure 4.4).

Each of the 3 plans (for 1, 2 and 3 beams) was performed with the same parameters, and different dosimetric values were recorded in order to evaluate the plans. Target mean doses for both, PTV and CTV and also $D_{2}$, $D_{98}$, $D_{95}$ were listed in Table 2. There is no big difference for maximum ($D_{2}$), minimum ($D_{98}$), $D_{95}$ and the mean doses between plans with 1, 2 or 3 beams. In any case, values did not differ more than 1Gy.

Table 2 Comparison of dosimetric parameters for the PTV and CTV

<table>
<thead>
<tr>
<th>PTV</th>
<th>CTV</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 beam</td>
<td>2 beams</td>
</tr>
<tr>
<td>$D_{2}$ [Gy]</td>
<td>78.50</td>
</tr>
<tr>
<td>$D_{98}$ [Gy]</td>
<td>74.20</td>
</tr>
<tr>
<td>$D_{95}$ [Gy]</td>
<td>76.80</td>
</tr>
<tr>
<td>$D_{\text{max}}$ [Gy]</td>
<td>77.80</td>
</tr>
</tbody>
</table>

$D_{x}$ = Dose which is delivered to the x% of the volume

Table 3 represents dosimetric parameters for OAR, were $V_{70}$, $V_{40}$, $V_{6}$ and $D_{\text{max}}$ are shown. Here also results are not very different from each other (less than 1Gy difference) for $V_{70}$ and $V_{40}$ in each plan. The maximum doses were also quite similar between plans. The only change was noticeable for small doses. The rectum volume which received 6% of the prescribed dose, for the plan with 3 beams, was more than 20% larger compared to plans with one and two beams.
Table 3  Comparison of dosimetric parameters for the rectum and bladder

<table>
<thead>
<tr>
<th></th>
<th>Rectum</th>
<th>Bladder</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>1 beam</td>
<td>2 beams</td>
</tr>
<tr>
<td>$V_{70}$ [%]</td>
<td>11.61</td>
<td>11.70</td>
</tr>
<tr>
<td>$V_{40}$ [%]</td>
<td>17.90</td>
<td>17.85</td>
</tr>
<tr>
<td>$V_{6}$ [%]</td>
<td>32.04</td>
<td>31.67</td>
</tr>
<tr>
<td>$D_{max}$ [Gy]</td>
<td>79.60</td>
<td>79.30</td>
</tr>
</tbody>
</table>

$V_x$: Volume which receive $x\%$ of prescribed dose

Dose distribution results are also seen on the DVH (Figure 4.5), where solid lines represent the plan for 3 beams, dashed lines for 2 beams and dotted for one beam. Plans for one and 2 beams have similar dose distribution and lines overlap each other, that is why on the DVH they cannot be clearly distinguished between dashed and dotted lines.

Figure 4.5 shows also an isodose curve comparison between the 3 plans. Even one beam is enough to achieve good target coverage, and well OAR sparing. The whole CTV volume (red line) as well as PTV (magenta line) is inside the 90% isodose (orange line). For plans with two and 3 beams this isodose is more adjusted to the tumor shape. The larger difference between those plans is visible in the dose to normal tissues. With one beam there is only one path through which the particles passes and it is covered almost entirely by the 70% isodose (yellow line), where for the 2 beam plan it is reduced to 50% (green line) and less, and for 3 beam plan to less than 30% (blue line).

5. Summary

The purpose of this report was to get a general knowledge on dose calculation, their main requirements and differences in available techniques, as well as explanation of first 3D dose computation performed for my study.

As far as advances in calculating the dose are concerned, from the times when first algorithms were implemented into TPS until present, there is still a need for developing faster and more accurate ones. Since calculation algorithm is the most crucial part of the TPS, its further development is essential for using hadron therapy in its full potential. As the computer speed and memories increase also the ability to use MC dose calculation for clinical routine is fast approaching and it may be the future for precisely planned particle treatment.
The first 3D dose computation for protons with pencil beam algorithm, using XiO, was explained with details. Each of the plans created highly conformal high dose region around the tumor. At the same time they delivered low doses to organs at risk. Also the integral dose to the patient was reduced which is the most important benefit that can be expected for pediatric patients.

This report was entirely dedicated to proton dose algorithms, mainly because the basic principles of the light ion dose computation are similar to the proton ones. However, the next deliverable ("D2. Publication on dose computation for light ions") will focus precisely on differences between proton and light ions dose computations.
Literature

8. S. Vatnitsky, Dose calculation in treatment planning of ion beam therapy. PARTNER treatment planning course. Vienna : s.n., February 2011.