Beam Modelling for Treatment Planning of Scanned Proton Beams

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Abstract

Scanned proton beams offer the possibility to take full advantage of the dose deposition properties of proton beams, i.e. the limited range and sharp peak at the end of the range, the Bragg peak. By actively scanning the proton beam, laterally by scanning magnets and longitudinally by shifting the energy, the position of the Bragg peak can be controlled in all three dimensions, thereby enabling high dose delivery to the target volume only. A typical scanned proton beam line consists of a pair of scanning magnets to perform the lateral beam scanning and possibly a range shifter and a multi-leaf collimator (MLC). Part of this thesis deals with the development of control, supervision and verification methods for the scanned proton beam line at the The Svedberg laboratory in Uppsala, Sweden.

Radiotherapy is preceded by treatment planning, where one of the main objectives is predicting the dose to the patient. The dose is calculated by a dose calculation engine and the accuracy of the results is of course dependent on the accuracy and sophistication of the transport and interaction models of the dose engine itself. But, for the dose distribution calculation to have any bearing on the reality, it needs to be started with relevant input in accordance with the beam that is emitted from the treatment machine. This input is provided by the beam model. As such, the beam model is the link between the reality (the treatment machine) and the treatment planning system. The beam model contains methods to characterise the treatment machine and provides the dose calculation with the reconstructed beam phase space, in some convenient representation. In order for a beam model to be applicable in a treatment planning system, its methods have to be general.

In this thesis, a beam model for a scanned proton beam is developed. The beam model contains models and descriptions of the beam modifying elements of a scanned proton beam line. Based on a well-defined set of generally applicable characterisation measurements, ten beam model parameters are extracted, describing the basic properties of the beam, i.e. the energy spectrum, the radial and the angular distributions and the nominal direction. Optional beam modifying elements such as a range shifter and an MLC are modelled by dedicated Monte Carlo calculation algorithms. The algorithm that describes the MLC contains a parameterisation of collimator scatter, in which the rather complex phase space of collimator scattered protons has been parameterised by a set of analytical functions.

Dose calculations based on the phase space reconstructed by the beam model are in good agreement with experimental data. This holds both for the dose distribution of the elementary pencil beam, reflecting the modelling of the basic properties of the scanned beam, as well as for complete calculations of collimated scanned fields.

Keywords: proton therapy, treatment planning, beam modelling, dose calculation, Monte Carlo

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## Contents

1 Introduction.......................................................................................................................... 9  
  1.1 The rationale and development of proton therapy ..................................................... 9  
  1.2 Treatment planning ........................................................................................................ 10  
  1.3 Scope and aim of this thesis ............................................................................................ 13  

2 Beam delivery techniques.................................................................................................... 14  
  2.1 Passively scattered beams ............................................................................................. 14  
  2.2 Actively scanned beams .................................................................................................. 15  
  2.3 Activation of collimator materials .................................................................................. 21  

3 Beam modelling................................................................................................................... 23  
  3.1 Beam model parameters ................................................................................................. 24  
  3.2 Beam model verification ................................................................................................. 26  

4 Collimator scattering .......................................................................................................... 28  
  4.1 Collimator scatter kernels .............................................................................................. 28  

5 Modelling of a scanned collimated proton beam.................................................................. 37  
  5.1 Collimator transport routine for an MLC ...................................................................... 37  
  5.2 Evaluation of the complete model ................................................................................ 40  

6 Conclusions and outlook...................................................................................................... 43  

7 Summary in Swedish ......................................................................................................... 45  

Acknowledgements............................................................................................................... 47  

References............................................................................................................................. 49
### Abbreviations

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>CAX</td>
<td>Central axis</td>
</tr>
<tr>
<td>CSD</td>
<td>Collimator to surface distance</td>
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<td>CSDA</td>
<td>Continuous slowing down approximation</td>
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<td>CT</td>
<td>Computed tomography</td>
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<td>FWHM</td>
<td>Full width half maximum</td>
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<td>IC</td>
<td>Ionisation chamber</td>
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<td>IMPT</td>
<td>Intensity modulated proton therapy</td>
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<td>MC</td>
<td>Monte Carlo</td>
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<td>MCS</td>
<td>Multiple Coulomb scattering</td>
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<td>MLC</td>
<td>Multi-leaf collimator</td>
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<td>MRI</td>
<td>Magnetic resonance imaging</td>
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<td>MWIC</td>
<td>Multi-wire ionisation chamber</td>
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<tr>
<td>PBK</td>
<td>Pencil beam kernel</td>
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<td>PET</td>
<td>Positron emission tomography</td>
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<td>SOBP</td>
<td>Spread-out Bragg peak</td>
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<tr>
<td>SSD</td>
<td>Source to surface distance</td>
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<tr>
<td>TIC</td>
<td>Transmission ionisation chamber</td>
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<td>TPS</td>
<td>Treatment planning system</td>
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<td>TSL</td>
<td>The Svedberg laboratory</td>
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1 Introduction

The science and technology of proton therapy is currently demonstrating an almost exponential growth. At the time of writing, a PubMed [1] search on “proton therapy” resulted in 418 papers published between 1977 and 2008, whereof 233 were published in the last five years and the yearly increase was approximately 14%. Proton therapy has developed from being a physics laboratory side activity with beam lines custom built piece by piece to full blown clinical facilities with commercial vendors offering turn key ready proton therapy centres.

The scope of proton therapy covers a number of disciplines, from radiobiology, cell biology and oncology to nuclear, accelerator and imaging physics and dosimetry and, not the least, computing science. This thesis deals with a specific part of the proton therapy science, namely the issue of how to model the beam emitted by a treatment machine as to accurately calculate the resulting absorbed dose for the purpose of planning optimal treatments.

1.1 The rationale and development of proton therapy

Proton and light ion beams have a very well defined range with the characteristic Bragg peak at the end, which distinguish these beams from the conventional photon beam modality. These properties make protons suitable to utilise for radiotherapy purposes, as first suggested by Wilson in 1946 [2]. Contrary to heavier particles, e.g. carbon ions which also are used for radiotherapy purposes, protons have a relative biological effect comparable to that of conventional modalities. The first treatment with a proton beam was performed in 1954 at Lawrence Berkeley Laboratory, USA [3-5]. These first proton therapy treatments targeted the pituitary gland and were performed with a 340 MeV beam, corresponding to a range of 63 cm in water, crossfiring through the patient [6]. Later lower beam energies were used to utilise the characteristic Bragg peak for therapeutic gain. During the following 15 year period, further development of proton therapy was performed at the Gustav Werner Institute (GWI) in Uppsala, Sweden, the Harvard Cyclotron Laboratory in Boston, USA, the Joint Institute for Nuclear Research in Dubna, Russia and the Institute of Theoretical and Experimental Physics in Moscow, Russia. The first treatment that targeted a
tumour was performed at the GWI, where the beam facility could yield both large field radiotherapy treatments and narrow beams for neurosurgical applications [7-12]. The team at the GWI was also the first to apply a range modulated Bragg peak to create a large uniform field as well as to achieve some of the first developments in scanned beam technology [13-15]. The first hospital based proton therapy facility, built at the Loma Linda University Medical Center, USA, started treatments in 1990 [16]. At present there are several clinical facilities in operation and a number under construction, mainly in Europe and the USA. The technique of shaping the dose distribution has been developed from large uniform fields to energy and intensity modulated proton therapy (IMPT), in which individual optimised weights are set to each pencil beam in order to create an optimal dose distribution that ensures high dose delivery to the target volume while sparing the healthy tissue.

A dedicated proton therapy facility is still very expensive and requires thorough investigations before investment. The decision to build a national proton therapy centre in Uppsala, Sweden, is based on several studies of the possible benefit of proton therapy for different tumour types [17-29].

1.2 Treatment planning

The delivery of radiotherapy is preceded by a treatment planning procedure. The planning starts by acquiring a set of images in which the target volume (tumour volume plus margins to account for tumour spread and position uncertainties) and sensitive structures can be defined and delineated. These images are most commonly acquired by means of computed tomography (CT). Less common but increasingly used are positron emission tomography (PET) and magnetic resonance imaging (MRI), to better map tumour cell prevalence and functional characteristics. Further, the irradiation fields are designed based on the defined structures and the desired/acceptable dose levels. The field properties such as beam angles, weights and apertures are set up iteratively, either manually or with the aid of automatic/semi-automatic optimisation algorithms. The goal of the treatment planning is to minimise the dose to healthy tissue to as large extent as possible while maximising the dose to the target volume(s), with the limitations given by the radiotherapy delivery equipment (hardware and software).

Since radiotherapy is delivered in fractions, the treatment planning procedure can also be adaptive and repeated during the progression of the treatment as to adapt to changes in patient geometry, this is called adaptive radiotherapy. The complete treatment planning procedure also includes more steps, such as plan evaluation, patient data handling, treatment scheduling etc, but these are beyond the scope of this thesis.
The dose calculation algorithms ("dose engine") used to support the treatment planning should be both accurate and fast. These two properties are often conflicting. The accuracy of the calculated dose depends on at least three important factors, the first of which is of course the accuracy of the interaction models and the particle transport algorithms of the dose calculation engine itself. The second factor is the representation of the patient anatomy, which consists of image data converted to material data via a pre-defined mapping. The topic of conversion from Hounsfield units (in the case of CT images) to material data has been subject to extensive research, both for conventional photon radiotherapy [30-34] and in particular for proton therapy [35-40] since inaccuracies in the material mapping for particles can lead to errors with respect to the beam range and Bragg peak location [41]. The third and last factor is the representation of the phase space of the particles emitted from the treatment machine. This latter part is facilitated by the beam model.

The beam model should include methods for characterisation of the treatment machine. Based on the characterisation, the beam model should provide fast reconstruction of the particle phase space to use as input to the dose calculation engine. The characterisation methods should be simple to understand, and result in a small, manageable amount of parameters with clear connections to the physical reality. If the beam model methods are based on measurements, these measurements have to be applicable in a clinical setting, i.e. simple and reliable to perform in the clinic on a routine basis. However, the beam model has to be complex enough to be able to reconstruct the phase space such as to fulfil the accuracy demands on the resulting dose distributions. This demand gets pronounced for more advanced beam delivery methods, such as scanned proton beams, since individual beam spots can have large weights compared to neighbouring spots. Inaccuracies in the description of individual beam spots will then not be compensated by neighbouring spots to the same rate as for a uniform field.

For proton beams there are two main methods for dose calculations, semi-analytical pencil beam kernels (PBK) methods using superposition of predefined dose data sets being one, and Monte Carlo (MC) methods using more explicit transport calculations of individual particles the other.

1.2.1 Pencil beam kernel methods for dose calculations
PBK methods [42-52] have long been the main dose calculation workhorse, mainly due to its simplicity and speed of calculation. PBK algorithms utilise dose deposition kernels describing the dose from a narrow pencil beam impinging onto semi-infinite slabs of water. The algorithm discretises the impinging beam into narrow pencil beams that are ray-traced voxel-by-voxel in the patient geometry along the beam direction. Scaling factors are
calculated for each depth and the final dose distribution is calculated by superposition of the correspondingly scaled pencil beam kernels. The proton transport processes can either be explicitly modelled by the scaling operations, or implicitly included into the pre-calculated kernels. The most important process considered is the widening of the beam due to multiple Coulomb scattering (MCS) and this is typically performed by applying a lateral Gaussian spread function. Some processes that may, or may not, be explicitly included are energy straggling and nuclear interactions. The PBK method superposes smooth analytical functions and yields noise-free dose distributions. However, to be efficient the incident phase space must be well behaved consisting of predominantly Gaussian distributions. The computation time scales more or less linearly with the number of fields and beam energies with no gain in computation time from calculation of overlapping fields.

1.2.2 Monte Carlo dose calculation methods

The basic idea of the MC method is to explicitly simulate the interactions of each particle as it passes through the matter, sampling the interactions based on the cross section for the actual interaction channels. However, protons undergo a vast number of interactions, yielding a problem far too immense to be simulated interaction by interaction for dose calculation purposes. The solution to this problem is the condensed history approach [53], where interactions of similar types are grouped together and sampled from a macroscopic cross section of the grouped interactions rather than from the microscopic cross section of each individual interaction. MC particle transport codes can be divided into two groups, class I and II. In class I codes the particle transport is performed with predetermined step lengths, with sampling of the interaction processes performed at the end of each step. The step length can be e.g. constant or logarithmic. In class II codes one group collectively only minor collisions in which the effect is small, and samples explicitly individual changes from major, “catastrophic” collisions. The distinction between minor and major collisions is usually determined by a threshold value, optionally set by the user.

MC codes capable of proton transport [54-59] has been less used in routine clinical treatment planning due to long calculation times compared to PBK. However, with the improvement of computer power and the introduction of faster MC algorithms [60, 61], MC can in the not so distant future be possible to use in a clinical system. MC offers many advantages compared to PBK methods. The dose distribution at heterogeneities can be more accurately calculated by MC methods and it is also easier to handle general particle distributions in MC simulations. One drawback of MC dose calculation is that it creates a noisy dose distribution. However, the statistics
are improved in regions with overlapping fields, which in the majority of treatment plans includes the target volume.

1.3 Scope and aim of this thesis

The scope of this thesis covers development and verification of dose delivery, beam modelling and beam line characterisation for scanned proton beams. The aim of the thesis is to:

- Develop control and verification methods for a scanned proton beam (I).
- Develop characterisation methods, from which beam line specific parameters can be extracted (II). These methods should be general and thereby applicable to different beam lines of varying geometry.
- Build a beam modelling framework upon which the phase space of a scanned proton beam can be constructed and used as input to a dose calculation engine (II).
- Study the distribution of protons out-scattered by a collimator and construct a calculation model to include this contribution into the calculation of the dose distribution in the patient (IV). In doing this it was also necessary to:
  - Investigate the ability of different Monte Carlo codes to transport protons at grazing incidence (III).

Although scanned proton beams have been experimentally established since 1980 [62], it is still a relatively unexplored area clinically. A considerable development, regarding practical and theoretical issues in both hardware and software will still be needed in order to fully exploit the clinical benefits of proton therapy with scanned beams. Regarding most of the work presented within this thesis, i.e. proton beam line characterisation methods and collimator scatter algorithms, preceding work published in the literature is scarce. The intension of this thesis is to take scanned proton beams one step further in clinical elaboration.
2 Beam delivery techniques

A single proton pencil beam cannot be efficiently used for therapy since a single Bragg peak is much too narrow both laterally and longitudinally to cover the target. The beam has to be spread out laterally and beams of several different energies have to be super-positioned in order to create a spread-out Bragg peak (SOBP) along the depth dimension. The lateral and the energy spreading can be performed both by passive scattering and active scanning of the proton beam. The passive scattering technique has the advantage of robustness and speed, while the active scanning technique has the potential of tailoring the high dose delivery exclusively to the target volume.

One important factor when considering different beam line designs is the neutron contamination of the beam [63-73]. In terms of beam line design, the more material in the beam, the more neutrons are generated that might hit the patient. Another radiation safety issue is activation of the beam line components, which is more of concern for the radiotherapy personnel rather than posing danger to the patient.

2.1 Passively scattered beams

The passive scattering technique, see Figure 1, is the most widely used technique today. The pencil beam delivered from the accelerator is widened to laterally cover the whole target by e.g. inserting a scattering foil into the beam. This creates a wide beam profile of Gaussian shape at the patient position, where the central part of the beam is approximately uniform. Optionally a second non-flat scattering foil, whose shape is optimised to create a flat beam profile can be used [74]. Flattening of the beam profile can also be achieved with concentric occluding rings [75] inserted into the beam downstream of the first scattering foil. The occluding rings completely block parts of the beam and scatter other parts of the beam into the blocked areas. Patient specific collimators are used to shape the lateral extension of the beam to the maximum lateral extension of the target volume. The energy modulation is, in order to create the SOBP, performed by using range modulator wheels [76] or ridge filters [77] customised to the maximum longitudinal extension of the target volume. Further, the dose distribution
can be shaped and adjusted in depth to the distal edge of the target by the use of range compensating filters. As shown in Figure 1, this beam delivery technique results in exposing some healthy tissue proximal to the target to high doses.

With the perspective of neutron contamination, it is clear that a passively scattered proton beam line contains a lot of material in which neutrons can be produced. The lateral spreading can however also be performed by a technique called wobbling, where the pencil beam is scanned in a predetermined pattern by scanning magnets. This latter technique takes us half way to actively scanned beams.

2.2 Actively scanned beams

A more refined technique is to use a so called actively scanned beam, where the Bragg peak of a narrow proton beam is scanned in three dimensions to cover the target. This technique enables tailoring the high dose delivery to the target volume only, see Figure 2. To fully take advantage of the flexibility of a scanned beam, one can make use of an optimisation algorithm and take the step to IMPT [78]. The first proton beam scanning system used
For patient treatment was developed at the Paul Scherrer Institute in Villingen, Switzerland [79].

For a scanned proton beam, the lateral spread of the beam can either be performed by the use of two scanning magnets [80-82] or one scanning magnet in combination with movement of the patient couch itself [79].

The longitudinal position of the Bragg peak is controlled by changing the energy of the beam. This is done either by absorber blocks or by actively choosing the extraction energy directly from the accelerator, which is only possible with a synchrotron. The range shifting absorber blocks can either be placed downstream of the scanning magnets or further upstream, closer to the accelerator. The downstream placement is the easiest to implement and control, a drawback is however that the lateral scattering in the absorber blocks will widen the beam creating a larger penumbra of the field. Absorbers will also increase the neutron contamination. These effects can be counteracted by placing the range shifter further upstream. The penumbra is then reinforced by downstream focusing elements. The neutron contamination can be reduced by subsequent bending of the beam and shielding. This technique puts a higher demand on the beam line downstream of the range shifter to be able to quickly adapt to different beam energies.

Optionally, a multi-leaf collimator (MLC) can be used to improve the lateral penumbra of the scanned field. The MLC will then need to be placed...
as close to the patient as possible, and will be adjusted to the largest lateral extent of each energy layer during treatment. An alternative to collimation for penumbra sharpening is spot weight optimisation with the objective to minimise the penumbra while maintaining a homogeneous dose distribution in the target volume. These two methods have been compared in a MC study by Bues et al [83], see Figure 3. The conclusion is that an MLC can indeed offer the possibility of a sharpened penumbra for shallow targets, while for deeper lying targets spot weight optimisation is a better choice. Safai et al [84] also reached the same conclusion in a comparison of the penumbra in fields produced by a collimated passively scattered beam and an un-collimated actively scanned beam. One should also keep in mind that an MLC will increase the neutron contamination of the field.

![Graph showing P_{80\%20} versus range R for different incident energies and spot weight optimisation vs MLC.](image)

*Figure 3.* The lateral penumbra, $P_{80\%20}$, versus range $R$, for three different incident energies, comparing the effect of spot weight optimisation of the penumbra to a collimation using an MLC. $P_{80\%20}$ is defined as the distance between the 80\% and 20\% isodose-linelines at the depth of the Bragg peak. The figure is adapted from Bues et al 2005.

### 2.2.1 Scanned proton beam development (I)

Part of this thesis deals with the development of the scanning system at the The Svedberg Laboratory (TSL) in Uppsala, Sweden [80]. In beam scanning with two magnets, the pole gap of the second magnet has to be wide enough to accommodate the beam width caused by the scanning of the first magnet. The size of the second pole gap limits the strength of its magnetic field, which leads to small deflection angles and thus a large source to surface distance (SSD). This results in very large systems with gantry diameters in the order of 10 m. The scanning system at the TSL represents an alternative solution to this problem. By mounting the second scanning magnet in a cradle which revolves around the geometrical centre of the first scanning magnet, see Figure 4, a small pole gap can be used in the second magnet allowing for shorter SSD. In order to minimise vibrations and mechanical...
stress on the system, the second magnet is moved continuously rather than in
discrete steps during irradiation. This scanning method results in a field
delivered in a raster pattern with slightly tilted scan lines. The energy is
changed with a range shifter consisting of slabs of PMMA, positioned as the
last beam modifying element in the beam line.

The scanning system is controlled by a program which runs in parallel
with an independent supervision system. The latter checks that all
parameters of the beam delivery are within pre-set boundaries, if not the
treatment is interrupted. The two systems get feedback from independent
detectors and sensors. During treatment, three transmission ionisation
chambers (TICs) and a multi-wire ionisation chamber (MWIC) are used to
monitor the dose delivered by the beam where the first TIC is used as the
monitor chamber and the following TICs and the MWIC are connected to
the supervision program, checking the dose delivery, the beam transmission
through the scanning magnets and the vertical position of the beam.

2.2.1.1 Dose verification and dosimetry methods
All beam delivery system for treatment of patients must be rigorously
checked and verified. The beam properties, such as beam profile and depth
dose, must be measured daily. The homogeneity of a uniform reference field
can be checked e.g. by using a fluorescent screen viewed with a CCD-
camera [85, 86]. Such a dosimetry system can also be used to measure the
relative dose distribution of a complex scanned field. However, the response
of the fluorescent screen is not linear with proton energy but shows a
quenching of the signal at lower energies. This means that the dose from
each scanned energy layer energy must be measured and corrected
individually by its own energy dependent quenching factor. The factors can
be retrieved experimentally as the ratio between a central axis (CAX) depth
Another dosimetry method providing the possibility of 3D dose verification is gel dosimetry. The dose deposited in a gel alters its chemical properties and these changes can be evaluated using MRI or optical scanners, depending on the type of gel. The types of gel reported to have been used for proton beams are ferrous sulphate gel [87], BANG gel [88] and normoxic polymer gel [89]. Gel dosimetry has been tested both at the passively scattered and the actively scanned proton beam line at the TSL, where for the latter the dose to a torus-shaped target volume was measured utilising normoxic methacrylic acid gel [90], see Figure 6. The gel was scanned after irradiation by MRI. One drawback of gel dosimetry is, as for the fluorescent screen, that the response is quenched for lower proton

Figure 5. CCD-system measurement of three layers in a scanned field with a spherical target volume. The fluorescent screen was positioned at a depth equal to the shallowest layer of the sphere. The upper left panel shows the relative intensity distribution from scanning the deepest layer, i.e. the highest energy. The upper right panel shows the relative intensity distribution from the central layer and the lower left panel shows the relative intensity distribution at the shallowest layer (lowest energy). A central profile through the sphere is shown in the lower right panel where all scanning layers of different depths have been corrected by the energy dependent quenching factor and then added.
energies. Another disadvantage is that this method requires scanning post irradiation making it rather time consuming. These drawbacks disqualify gel dosimetry for daily use, e.g. for plan verification. The method can however be very useful in less frequent quality assurance measurements, such as commissioning.

In a MC dose calculation, the quenching of the response from the fluorescent screen or the gel can be taken into account in two ways, either by calculating the expected fluorescence/chemical change by correcting the deposited dose for each step of the transportation by means of an experimentally acquired quenching factor. The fluorescence/chemical change process can also be simulated directly in detail along the lines as has been done for luminescence by Edmund et al [91].

2.2.1.2 Dose-per-monitor-unit formalism and normalisation (II)

The prescribed dose calculated by the TPS has to be converted to monitor units, which in general is a signal proportional to the number of particles
traversing the monitor chamber. In practice, a formalism linking monitor units to dose under reference conditions is needed. This type of formalism guarantees exact reconstruction of the measured calibration dose by the calculation for reference conditions. Contrary to the current code of practice for dosimetry for proton beams [92], the work in this thesis suggests that the reference conditions should be; the calibration measurement is to be performed in the plateau region of the depth dose curve, at a depth of 4.5 cm in water. Further, the reference measurement should be performed for a mono-energetic field with a range shifter block of 3.0 cm PMMA in the beam.

2.3 Activation of collimator materials

In order to estimate the magnitude of the radiation safety issue caused by the induced activation of collimators, blocks of brass (used for collimators in passively scattered beams) and a tungsten alloy (95% tungsten, 3.5% nickel and 1.5% copper, likely to be used in MLCs for actively scanned beams) were irradiated for 10 min with a 180 MeV proton beam of 1 nA beam current. The resulting activity was measured with a portable dose-rate meter (RNI 10/R). The thickness of the two blocks in the direction of the beam was 5 cm and 4.5 cm for the tungsten and brass block respectively. The range of 180 MeV protons in the tungsten alloy is 2.3 cm and in brass 3.9 cm, i.e. the beam was completely absorbed by both blocks. The activity was measured at given time intervals up to 12 h after irradiation at three positions, at the distal surface, 30 cm downstream of the distal surface and at the proximal surface as shown in Figure 7.

The measurement position most relevant when considering the radiation safety of the radiotherapy personnel is position (b) in Figure 7, 30 cm downstream of the distal surface of the collimator block. The staff will most likely be in this area when setting up patients for treatment. The dose limits set by the Swedish radiation protection authority for people working with ionising radiation is 50 mSv/year for an individual year, and 100 mSv in total for five consecutive years [93] while the limit to the extremities is set to 500 mSv/year. The level of activity 5-10 min after irradiation, the time interval when the staff is likely to be in the room and in the vicinity of the collimator, was in this experiment 150-200 μSv/h for the brass and 30-60 μSv/h for the tungsten. Supposing the technicians spend 5 min in the vicinity of the collimator per patient, with 10 patients per day and 48 weeks of work per year this adds up to 200 h yielding 30-40 mSv/year for the brass and 6-12 mSv/year for the tungsten of the experiment. The major part of this dose will probably be deposited in the extremities, especially regarding the patient specific brass collimators, which are positioned manually. On the other hand, an MLC made of tungsten will remain in place and get a renewed
boost of activation for every treatment. The mean whole body dose to people working with radiotherapy (oncologists, nurses and hospital physicists) in Sweden was in 2005 in the order of 0.5 mSv/year [94]. The mean dose to the hands for hospital physicists was in the same year 17 mSv/year. Although the geometrical circumstances and level of irradiation of a real collimator will differ from this experiment, it shows that a considerable part of the yearly radiation exposure to proton radiotherapy personnel will be from the induced activity of the collimator. However, efficient logistic procedures for patient setup can significantly reduce the exposure time and limit the staff dose to levels achieved in conventional therapy.

![Diagram](image)

**Figure 7.** Measurement of the induced activity versus time after irradiation of two collimator blocks. The exposure time was 10 min with a 180 MeV proton beam of 1 nA beam current with the beam completely absorbed by the block. The positions of the measurements are pictured in upper left panel.
3 Beam modelling (II)

The dose calculation needs a specification of the particle energy, direction and position, i.e. the phase space, of the incident beam. This can be facilitated by a beam model. The beam model should provide the phase space such that the dose calculations can be done with enough detail and accuracy. Beam modelling and beam phase space characterisation methods have been extensively developed for electrons and photons [95-97], but for protons this area is relatively unexplored. Recently Schaffner [52] published a work of beam modelling with the aim of generality, otherwise only a few characterisations for individual machines have been published [98, 99].

Beam modelling can be performed in different ways. Detailed MC simulation of the complete particle transport through the treatment machine is at one end of the spectrum [100], this is an explicit method which results

![Diagram of beam delivery system and treatment planning system](image)

*Figure 8. The role of the beam model in the perspective of treatment planning and beam line characterisation.*
in large phase space files. Detailed simulation requires complete and detailed blueprints of the treatment machine, something which is not always available or even accurate. In this thesis, the goal is to develop a general beam model, which can be applied to a variety of beam lines. This beam model is based on a standardised set of measurements, which together with a limited set of geometrical data and a priori pre-calculated data, models the beam as composed of explicit, parameterised sources. The model is controlled by parameters which can be adjusted to specific proton beam line designs. The parameters are extracted by optimisation towards the measured data.

The role and tasks of the beam model are pictured in Figure 8. The characterisation process results in a set of beam model parameters, which are stored in the TPS database. As shown in Figure 8, the characterisation methods of the beam model set up the interface between the TPS and the treatment machine. When performing a dose calculation, beam model parameters and machine specifications are retrieved from the database and used according to algorithms defined by the beam model to construct the beam phase space. The beam modelling methods developed within this thesis have focused on scanned proton beams, and have been thoroughly tested for the scanned beam at the TSL.

3.1 Beam model parameters

All parameters of the beam model are effective parameters, i.e. parameters are assigned values that reconstruct measurements and not necessarily represent the “true” values or distributions. The parameters describe the direction, the radial and angular distribution and the energy spectrum of the incident pencil beam. The coordinate system used in the following sections has the $x$- and $y$-axis as the lateral dimensions and the $z$-axis is oriented along the un-deflected beam direction.

3.1.1 Pencil beam direction

For a scanned beam, the direction of the pencil beam is set by the scanning magnets. This is described by applying a rotation matrix, $\mathbf{R}$, for each deflection by angles $\theta_x$ and $\theta_y$ to the direction vector $\mathbf{v}$, thereby calculating the new direction, $\mathbf{v}'$, as

$$\mathbf{v}' = \mathbf{R}_y(\theta_y)\mathbf{R}_x(\theta_x)\mathbf{v}.$$  \hspace{1cm} (1)

In order to calculate the rotation matrix, the axis of the rotation has to be determined as well as the rotation angle. The rotation angle is determined by a focal point, $f_x$ and $f_y$, for the scanning in each direction, i.e.
\( \theta_x = \theta_x(f_x, f_y, r) \) and \( \theta_y = \theta_y(f_x, f_y, r) \) where \( r \) is the position of the proton in a reference plane. The focal points are nominally located at the centre of each scanning magnet. However, the position of the effective focal points used in the beam model will in general differ slightly from the nominal positions due to imperfections in beam optics, etc. The effective focal points are therefore extracted from measurements of the beam position at two planes perpendicular to the beam for a number of different beam deflections. These measurements are performed in air with a Hi-pSi diode [101] mounted on a 3D-servo.

### 3.1.2 Radial and angular distributions

The radial and angular distributions of the elementary pencil beam are assumed to be approximately Gaussian, i.e. the distribution of protons in the \( x \)-component, \( \Phi_x \), can be written as

\[
\Phi_x(\alpha, x) = \frac{\exp \left( -\frac{\sigma_a^2 x^2 - 2\sigma_{xa} x \alpha + \sigma_x^2 \alpha^2}{\sigma_a^2 \sigma_x^2 - \sigma_{xa}^2} \right)}{\pi \sqrt{\sigma_a^2 \sigma_x^2 - \sigma_{xa}^2}}, \tag{2}
\]

where \( \alpha \) is the angle between the \( z \)-axis of the beam and the projection of the direction in the \( xz \)-plane of the beam. The parameters of the Gaussian, i.e. the mean square angular spread \( \sigma_a^2(z_0) \), the covariance \( \sigma_{xa}(z_0) \) and the mean square radial spread \( \sigma_x^2(z_0) \) are determined at a starting plane, \( z_0 \), by optimisation towards measured beam profiles. These profiles are measured with a fluorescent screen/CCD camera set-up [85].

### 3.1.3 Energy spectrum

The energy spectrum specifies the distribution of energies for the non-modulated pristine beam. Rather than measuring the energy spectrum, \( \Phi_E(E) \) directly for a pencil beam, a measured CAX depth dose curve of a large scanned field is used to extract an effective energy spectrum. Calculated depth dose curves based on mono-energetic protons are superpositioned and weighted by a Gaussian,

\[
\Phi_E(E) = \frac{1}{\sqrt{2\pi} \sigma_E} \exp \left( -\frac{(E - E_0)^2}{2\sigma_E^2} \right). \tag{3}
\]
The mean energy $E_0$ and standard deviation $\sigma_E$ are optimised for the superpositioned depth dose curve to fit to a depth dose curve measured in water with a Hi-pSi diode [101] mounted on a 2D-servo. This method has the advantage that it ensures that the calculated depth dose has the same range as the measured one. The depth penetration of the calculated dose will then be independent of the accuracy of the stopping power tables used in the dose engine (as long as the dose engine is using consistent stopping power tables). For treatment planning purposes this is preferred as the range is more important than an exact determination of the beam energy.

### 3.1.4 Energy shifting

As mentioned in section 2.2, the longitudinal position of the Bragg peak is controlled by shifting the energy. For beam lines that use active selection of the extraction energy, or use a range shifter placed upstream close to the accelerator, the beam modelling methods to determine the radial and the angular distributions and the energy spectrum described in section 3.1.2 and 3.1.3 have to be repeated for a set of energies emitted from the accelerator.

If the range shifter is placed downstream of the scanning magnets it does not only decrease the energy of the beam, the energy spectrum is also widened by energy straggling. Further, the radial and angular distributions are widened due to multiple scattering, primary protons are absorbed and secondary protons are produced due to inelastic hadronic reactions. Due to the wide range of effects that this type of range shifting has on the beam, the energy shifting is handled by the beam model separately through use of a dedicated proton MC transport step, modelling the above mentioned interactions. Secondary electrons and nuclear fragments created in the range shifter are assumed to have too short range to escape from the range shifter and their energy is deposited locally.

### 3.2 Beam model verification

In this thesis the beam model has been verified against measured beam profiles and CAX depth dose distributions for different energies. A crucial test for the beam model is to construct the phase space for an elementary beam, i.e. an individual pencil beam, since this constitutes the building block of a field delivered by a scanned proton beam. For this purpose, the dose delivered by four different stationary elementary beams was measured at the scanned proton beam at the TSL in a water phantom with a Hi-pSi diode [101] mounted on a 3D-servo. Two of the elementary beams had the full energy, one directed straight forward and the other deflected by both
scanning magnets. The other beams had the energy reduced by insertion of 5.0 cm of PMMA in the range shifter. As for the non-modulated beam, one straight forward and one deflected elementary beam were used. For each elementary beam, the dose distribution in five planes was measured. The corresponding phase space for each elementary beam was constructed by the beam model and the general purpose MC code GEANT 3.21 [55] was utilised as dose engine to calculate the dose, see Figure 9. The resulting sets of measured and calculated dose distributions were compared by using the gamma index [102] where 96.5 % of the dose points in total passed the gamma 1 % / 1 mm criterion and 99.8 % passed the gamma 2 % / 2 mm criterion.

Figure 9. Isodose curves for the four elementary beams, (a) full energy no deflection, (b) full energy deflected by both scanning magnets, (c) range shifted with 5.0 cm PMMA no deflection and (d) range shifted with 5.0 cm PMMA deflected by both scanning magnets. The solid lines are the measurement and the dashed lines are the calculation based on the phase space constructed by the beam model.
Collimation close to the patient surface is always used for passively scattered beams to limit the field to the lateral extent of the target, as shown in Figure 1. Although the high dose distribution can be better confined to the target volume using actively scanned beams, collimators can still be used to sharpen the lateral penumbra as mentioned in section 2.2.

However, collimators also generate scattered particles. The scattered beam component can, depending on the aperture shape and size and incident beam phase space, contribute with up to 15% of the dose locally [103, 104]. For accuracy reasons it is therefore highly desirable to include the contributions from collimator scatter in the dose calculation of the TPS. Due to the complex phase space of collimator scatter, it is difficult to model within the limitations of current PBK dose engines and has therefore usually been neglected [44, 105, 51, 106, 107, 52].

Published work on collimator scatter mostly report evaluations of total dose distributions for collimated fields or studies of the phase space of collimator scatter protons in bulk [108, 103, 109, 104]. In this thesis, a calculation model for collimator scatter based on parameterisation of pre-calculated scatter kernels is presented, analogous to what is used for photons [110] and electrons [111]. The parameterisation is based on an in-depth study of the characteristics of collimator scatter. The calculation model presented here is designed to provide the phase space of collimator scatter by sampling, i.e. it does not directly address the problem of implementing collimator scatter into a PBK dose engine. However, based on the understanding of the constituents of collimator scatter offered by this model, it might also be used for solving the PBK-collimator scatter problem.

4.1 Collimator scatter kernels

A collimator scatter kernel describes the distribution of protons out-scattered from a collimator element irradiated by a point mono-directional beam. The kernels in this work have been pre-calculated by MC, as described in section 4.1.2. The collimator scatter can be divided into two categories, inner and front face scatter, based on the incidence conditions, see Figure 10. The geometrical conditions for both inner and front face scatter are very demanding for the condensed history transport mechanics applied by the MC
code, i.e. particles travelling very close to, and nearly in parallel with a lateral boundary with deflections dominated by multiple Coulomb scattering (MCS). Hence, the ability of the MC code to produce correct results under these circumstances has to be tested and verified towards experimental data. Ideally, both the MC transport for inner and front face incidence should be tested, but out-scatter following inner face impact is very hard to isolate and measure separately. However, the proton tracks inside the collimator following both inner and front face incidence have similar properties. Thus, investigating the particle transport at front face incidence also provides insights into the quality of transport for inner face incidence.

4.1.1 Experimental test of MC particle transport at grazing incidence (III)

In order to test MC transport at grazing/front face incidence, the out-scatter following front face impact has been simulated with the MC codes GEANT4.8.1, GEANT4.8.2 [58, 112], FLUKA2006 [113, 59] and MCNPX2.4.0 [57]. The simulation results have been compared to experimental data measured at the TSL.

Adequate modelling of MCS is essential for accurate transport modelling under grazing incidence. Unfortunately, the degree of sophistication regarding modelling MCS achieved in electron MC [114], has no counterpart.
in current MC codes transporting heavier particles, including the codes tested in this work [115].

In order to test the transport at grazing incidence the calculated dose deposited by protons out-scattered following grazing incidence on a large scatter block, to a thin film positioned downstream of the scatter block was compared to measurements, see Figure 11. The geometry of the simulations and the experiments consisted of a narrow beam impinging on a large (in comparison to the beam size) tungsten block, thus mimicking front face incidence. Four different incidence angles, $\theta_{\text{in}} = 3.5, 5.0, 7.5$ and $10.0^\circ$ and two incident energies, $E_{\text{in}} = 180$ MeV and 98 MeV were used.

4.1.1.1 Simulation set-up

In the MC codes studied in this work, the user has more or less influence on the particle transport. One important aspect of the study was to determine the impact of user settings on the distribution of out-scattered protons following grazing incidence. The parameters judged to have impact on MCS modelling and/or transport stepping were varied in the simulations. This subset of parameters included control of particle transport in thin layers, backscattering and transport step length (GEANT4), the MCS modelling threshold (FLUKA) and the energy transport cut off value (MCNPX).

4.1.1.2 Experimental set-up

Experimentally, out-scatter following grazing incidence was measured with the fluorescent screen and CCD-camera set-up described in section 2.2.1.1. A narrow beam was created by a collimator arrangement in two parts. The first collimator consisted of perpendicular tungsten slits, creating a square aperture of 0.3 cm size. The second collimator was introduced to shield off protons scattered by the first collimator. The second collimator, made of brass, had a circular aperture of 0.8 cm diameter. Following the collimators
was a tilted tungsten block, hit by the narrow collimated beam at a small angle. The fluorescent screen was placed as close as possible to the tilted block, in order to cover as large a solid out-scatter angle as possible. In order to compensate for the energy dependent quenching of the light signal from the fluorescent screen/CCD-camera system, an experimental quenching factor was deduced as described in section 2.2.1.1. The measurements were compensated for the quenching by applying a correction, calculated by MC utilising the experimental quenching factor.

4.1.1.3 Results
The study of the impact of the setting of the simulation parameters showed the importance of an accurate and careful set-up of transport parameters in the MC simulations. Further, the results of the GEANT4 simulations contained un-physical transport artefacts, including an asymmetry in the out-scatter fluence and discontinuity of the derivative of the out-scattered energy spectrum. The latter can be attributed to the lack of an explicit boundary crossing algorithm, which adapts the transport while approaching a boundary.

However, the study showed that all codes were capable of calculating out-scatter distributions, see Figure 12, in good agreement with measurements, provided a careful set-up of the transport parameters, as shown in Figure 13.

Figure 12. Isodose plots for the dose distribution for $\theta_{\text{in}} = 3.5^\circ$, for $E_{\text{in}} = 98$ MeV in the left panel, and for $E_{\text{in}} = 180$ MeV in the right panel. In both panels, the solid lines represent the experiment, corrected for the quenching and the dashed lines represent the GEANT4.8.1 simulation. Isodose lines are drawn in 10%-steps of the maximum dose where the 10 and 50% levels have been labelled.
Figure 13. Comparison of simulations with the “optimum” transport parameter settings to experiment for $E_{in} = 98$ MeV (left column) and $E_{in} = 180$ MeV (right column). The upper panel row shows the dose weighted mean emission angle, $\bar{\nu}$, the middle panel row shows the total solid angle for which $D > 0.5D_{max}$, $\Omega_{50}$, and the lower panel row shows the dose weighted out-scatter probability, $P_D$. 
4.1.2 Calculation and parameterisation of scatter kernels (IV)

As the contribution of collimator scatter is rather small in comparison to the total dose, some approximations in the kernel parameterisation can be made without jeopardising the overall dose calculation accuracy. The goal in this work was to obtain a compact representation of the collimator scatter kernels that enables a sufficiently accurate determination of scatter dose for collimated fields.

Utilising the knowledge in optimal particle transport parameter settings, gained from the comparisons described in section 4.1.1, collimator scatter kernels were calculated by the MC code GEANT4.8.2, by scoring out-scattered protons differential in energy and direction, following incidence of a pencil beam (mono-energetic, mono-directional point source) upon either the inner or the front face of a large collimating block. In this way, scatter kernels for inner face impact were generated for specified distances to the collimator edge, $d$, in the interval $0.001$ cm to $0.5$ cm and for front face impact, for incidence angles, $\theta_{\text{in}}$, in the interval $0.01^\circ$ to $10.0^\circ$, see Figure 10. Two incidence energies were used, $E_{\text{in}} = 180$ MeV and $98$ MeV. The material used in this study was a tungsten alloy (95% tungsten, 3.5% nickel and 1.5% copper, density $18.0$ g/cm$^3$). All together this resulted in a data set representing 54 different out-scatter distributions, differential in energy and direction. The aim was to parameterise this data set, i.e. to squeeze the fat bulging triple differential distribution into a rather slim neoprene suit of analytical functions. The parameterisation was aimed at facilitating sampling. Starting from the scored fluence distribution $f(x)$, the cumulative probability, $P(X)$, was calculated as

$$ P(X) = \int_{a}^{b} f(x) \, dx $$

(4)

where $a<x<b$ and $\int_{a}^{b} f(x) \, dx = 1$. The function of interest in the parameterisation was the inverse

$$ X = P^{-1}(u) $$

(5)

in which $u$ is in the interval 0-1. Thus, generating random numbers, $u$, uniformly distributed between zero and one, i.e. $u \in U(0,1)$ enables sampling from the desired function $f(x)$.

The function corresponding to $P^{-1}(u)$ was calculated from the scored distribution of out-scattered protons. In the search of possible short-cuts in the parameterisation, it was found that the phase space distributions were
approximately independent of the incident energy. This was taken advantage of by grouping the inner face out-scatter by \( d_R = d/R \), where \( R \) is the continuous slowing down approximation (CSDA) range, and scaling the energy spectra to the incident energy \( E_{\text{in}} \). The chosen sampling order started with the out-scatter probability, \( P_S \), which was found to be nicely fitted by the exponential function,

\[
P_S(b) = \exp \left( s_1 b^2 + s_2 b + s_3 \right)
\]  

(6)

where \( b \) represents the impact parameter \( d_R \) or \( \theta_{\text{in}} \), depending on the scatter face (inner or front), and \( s_i \) are the fitted parameters. The second step was to sample the azimuth angle, \( \varphi_{\text{out}} \), for which the sampling function was derived as

\[
\varphi_{\text{out}}(b, u_1) = p_1 u_1^2 + p_2 u_1 + p_3 \left( \exp(p_4 u_1) - 1 \right),
\]

(7)

where \( p_i = p_i(b) \) are the fitted parameters. The third variable to be sampled was the polar angle, \( \theta_{\text{out}} \), generated from

\[
\theta_{\text{out}}(b, \varphi_{\text{out}}, u_2) = r_1 u_2^3 + \exp(r_3 u_2^3 + r_4 u_2^2 + r_5 u_2 + r_6),
\]

(8)

where \( r_i = r_i(b, \varphi_{\text{out}}) \) are the fitted parameters. In this way the covariance between \( \varphi_{\text{out}} \) and \( \theta_{\text{out}} \) was incorporated into the calculation model. The fourth and last quantity to be sampled was the out-scatter energy, \( E_{\text{out}} \). In the parameterisation of the out-scatter energy the covariance between the azimuth angle and the energy was ignored as an approximation and the energy was parameterised differential in polar angle only with the function

\[
E_{\text{out}}(b, \theta_{\text{out}}, u_3) = E_{\text{in}} \left( q_1 u_3^{1/2} + q_2 u_3^{1/3} + q_3 u_3^{1/4} \right)
\]

\[
+ \exp \left( q_4 u_3^3 + q_5 u_3^2 + q_6 u_3 \right) - 1,
\]

(9)

where \( q_i = q_i(b, \theta_{\text{out}}) \) are the fitted parameters. Thus, sampling four independent uniformly distributed random numbers, \( u_i \in U(0,1), i = 1-4 \), and applying the equations (6)-(9) generates a proton out-scattered from a collimator.
4.1.3 Verifying the parameterisation

Since the endpoint of the parameterisation is to be able to calculate the dose contribution from collimator scattered protons to the total dose distribution, dose deposition was chosen to be the basis of verification of the parameterisation. Thus, as a first test, two sets of MC calculations of the dose distribution in water for a set of typical values for the impact parameters were compared. The first set utilised the parameterisation as input and the second set utilised the MC generated phase space file from the kernel generation as input, see Figure 14. These two sets of dose distributions have been compared by means of the gamma index [102] where in total 82.1%, 96.7% and 99.3% of the voxels fulfilled gamma $x$ % / $x$ mm for $x$ = 1.0, 2.0 and 3.0 respectively.
Figure 14. Dose distributions for inner and front face out-scatter kernels corresponding to $d_R = 0.085$ and $\theta_{in} = 3.0^\circ$, respectively. The left column is the full MC calculation based on the MC generated phase space file and the right column is based on the parameterisation. The dose distributions have been normalised to the maximum dose of the full MC simulation, $D_{\text{max}}^{\text{G4}}(d_R, E_{in})$ and $D_{\text{max}}^{\text{G4}}(\theta_{in}, E_{in})$ respectively.
5 Modelling of a scanned collimated proton beam (II+IV)

As shown in Figure 2, the main beam modifying elements included in a scanned proton beam line are the scanning magnets, the optional range shifter and an optional patient specific aperture realised by an MLC. The first two elements, the scanning magnets and the range shifter are, together with the basic properties of the scanned pencil beam modelled by the beam model described in section 3. In order to account for a variable collimation device, such as the MLC as shown in Figure 2, the model was extended with a collimator transport routine, in which individual protons are ray-traced through a model of the MLC by a dedicated MC code that samples collimator scatter from the parameterisation described in section 4.1.2.

5.1 Collimator transport routine for an MLC

The purpose of the collimator transport routine is to transport protons and generate possible collimator scatter for an MLC aperture, as defined by the leaf positions. The protons are created at a source plane, located upstream of the MLC, by the beam model according to the current fluence distribution and transported downstream to the upstream plane of the MLC where the collimator transport routine takes over. In the current implementation of the routine only MLC geometries with rectilinear leaves (i.e. non-focussed sides and leaf tips) are considered. The transport routine proceeds by tracing the protons through the MLC geometry. Since the thickness of an MLC is small, typically less than 10 cm, all interactions with air is neglected. For each proton, the collimator transport routine moves the proton along its initial direction to the intersection with any leaf surface, or to the downstream plane of the MLC, whichever comes first. If the proton intersects with a leaf surface, see Figure 15, the impact parameters are calculated, i.e. the distance to the edge divided by the CSDA range, $d_R$, in the case of inner face impact and the incidence angle, $\theta_{\text{in}}$, in the case of front face impact. Based on the impact parameter and the incidence energy, $E_{\text{in}}$, out-scatter is sampled using equation (6)-(9). If a proton is out-scattered, it is displaced a distance $\Delta R = R(E_{\text{in}}) - R(E_{\text{out}})$ along the projection of its incidence direction on the out-scatter surface. The lateral displacement due to multiple scattering is
neglected. If the proton after displacement ends up inside another leaf, the proton is discarded. On the contrary, if it ends up in mid-air by exiting the leaf laterally, no correction is made. Sampling of out-scatter is only performed for protons with impact parameters that are within the boundaries of the kernel parameterisation, i.e. in the case of inner face impact for \[ 0 \leq d_R \leq 0.21 \] and in the case of front face impact for \[ 0^\circ \leq \theta_{\text{in}} \leq 10.0^\circ \]. At both of the upper boundaries the out-scatter probability is 5% or less.

### 5.1.1 Inner face impact

For inner face impact, the routine must determine through which front face surface the out-scatter proton will exit. For this purpose, the distances to all

![Schematic picture of four different cases of collimator impact](image)

**Figure 15.** Schematic picture of four different cases of collimator impact, with the arrows indicating the impact and out-scatter directions and locations. The beam direction is from left to right.

- **Case 1:** Ordinary inner face incidence, the out-scatter surface is chosen as the closest open surface, i.e. not closed by a neighbouring leaf, as indicated by the dotted lines on the inner face of the leaves.
- **Case 2:** Ordinary front face scatter. The effect of neglecting the limited extension of the leaf surfaces can have the effect that protons are displaced into mid-air, as indicated by the lower arrow.
- **Case 3:** Punch through to the downstream plane of the collimator.
- **Case 4:** Special cases, the upper impact shows a double scatter event in which the out-scattered proton impacts a second time on the front face of the collimator. The lower arrow shows a proton that after displacement ends up inside a neighbouring leaf and is discarded.
open edges of the leaf of impact are calculated. An edge is considered open if the neighbouring leaf, in the perpendicular direction of the leaf movement, is retracted beyond the point of impact or if the two opposing leaves, in the direction of the leaf movement, are retracted from each other. Out-scatter is sampled based on the shortest distance only and the corresponding leaf surface is chosen as the out-scatter surface.

With a realistic incident beam phase space and MLC geometry, the incidence on the inner face of the MLC will in most cases differ from normal incidence, which was the circumstance in the kernel generation and parameterisation. Both the out-scatter probability and the phase space distribution of out-scattered protons depend on the distance to the edge and the angle of incidence. In order to compensate for this discrepancy between the parameterisation and reality, the out-scatter probability was calculated, by means of MC simulations, both as a function of \( d_R \) and

\[
\gamma_p = \arccos \left( \hat{v}_{\text{in}} \cdot \hat{v}_{\text{proj}} \right),
\]

i.e. the angle between the incidence direction and its projection on the out-scatter surface. In the collimator transport routine, out-scatter is sampled from \( P_S \left( d_R, \gamma_p \right) \) while the out-scatter direction and energy distributions are unchanged.

### 5.1.2 Front face impact

For front face impact, the implementation is rather straightforward. The only case that needs special attention is possible punch through, i.e. when the proton exits through the downstream surface of the MLC instead of through the front face. This may happen when the impact occurs within a distance from the exit plane of the collimator which is shorter than the range of the proton. This means that the sampling from equation (6)-(9) has to be adjusted to take the limited longitudinal extension of the collimator leaf into account. In the current implementation, this is handled by calculating the range for each proton out-scattered from the front face, before and after the scattering event. If the change in range is larger than the distance from impact point to the exit plane (in the incidence direction of the proton) the sampled out-scatter direction and energy are discarded and the proton is treated as a possible punch through event.

### 5.1.3 Punch through

The transport of possible punch through protons is handled by a dedicated MC algorithm, which includes multiple scattering, energy loss and absorption. The lateral displacement and angular deflection caused by multiple scattering is sampled at the exit plane by applying the approximations of the Fermi-Eyges transport equations suggested by Russell
et al [116]. The energy loss is calculated by look-up in stopping power tables, extracted from GEANT4.8.2, based on the reduction of the range. Energy straggling is neglected. The absorption due to inelastic nuclear reactions is sampled from an energy independent approximation of the macroscopic absorption cross section.

5.2 Evaluation of the complete model

The collimator transport routine incorporated into the beam model was evaluated by comparison of simulated dose distributions in a water phantom for a set of MLC geometries. Based on the phase space created by the beam model, dose deposition in a water phantom was calculated by the MC code GEANT4.8.2. Two sets of MC calculations were performed, where the phase space for the first set was created downstream of the collimator. The second set of simulations was based on the phase space extracted at a plane upstream of the collimator with the MLC included in the GEANT4 simulation geometry, thus letting GEANT4 perform the particle transport in the MLC. Finally, the dose distributions were also compared to experimental data for a few geometries.

5.2.1 Simulations

Two different MLC geometries were considered where the first was a simplified model of an MLC with three leaf pairs, corresponding to the collimator set-up of the experiments described in section 5.2.2. The MLC material in this implementation of the model is the same tungsten alloy used in the generation of the scatter kernels, i.e. 95% tungsten, 3.5 % nickel and 1.5% copper, with a density of 18.0 g/cm³. The leaves of this MLC measured 8.6 cm in width and 5.0 cm in thickness (along the beam axis). The collimator to surface distance (CSD) was set to 5.0 cm for all simulations when not stated otherwise. Three different rectangular apertures were simulated, measuring $2.0 \times 8.6$, $4.0 \times 8.6$ and $8.6 \times 8.6$ cm². The irradiation field was a homogeneous, mono-energetic field of two different energies, 180 MeV and 98 MeV. The lateral extension of the field was large enough to irradiate the whole perimeter of the aperture including a margin. For the aperture measuring $4.0 \times 8.6$ cm² the CSD was varied in the interval 2.1 - 40.0 cm.

The second MLC geometry simulated a more realistic MLC, consisting of 80 leaf pairs of 0.5 cm width and 5.0 cm thickness. This MLC was used to realise two different apertures, a tilted square and a comb shaped aperture. These apertures constitute a hard test for the collimator transport by exposing large areas of the inner and front face to the incident field. The incidence energies in these simulations were set to 180 MeV and 150 MeV.
Table 1. The result of the gamma index comparison of the dose distribution based on the parameterised phase space versus the full MC simulation. The result is given per incident energy for all apertures, in total for the rectangular apertures and individually for the comb and tilted square apertures.

<table>
<thead>
<tr>
<th>Aperture</th>
<th>$E_{in}$ (MeV)</th>
<th>Fraction of dose points passing gamma 1% / 1 mm (%)</th>
<th>Maximum deviation (gamma x % / x mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rectangular</td>
<td>180</td>
<td>99.4 - 99.5 - 99.6</td>
<td>2.6</td>
</tr>
<tr>
<td>Rectangular</td>
<td>98</td>
<td>98.4 - 98.9 - 99.4</td>
<td>1.9</td>
</tr>
<tr>
<td>Comb</td>
<td>180</td>
<td>97.5</td>
<td>2.3</td>
</tr>
<tr>
<td>Comb</td>
<td>150</td>
<td>99.2</td>
<td>1.9</td>
</tr>
<tr>
<td>Tilted square</td>
<td>180</td>
<td>99.1</td>
<td>2.1</td>
</tr>
<tr>
<td>Tilted square</td>
<td>150</td>
<td>99.1</td>
<td>2.2</td>
</tr>
<tr>
<td>Total</td>
<td>-</td>
<td>99.1</td>
<td>2.6</td>
</tr>
</tbody>
</table>

5.2.2 Experiments
The three leaf pair geometry described in section 5.2.1 was realised experimentally by stacking blocks of a tungsten alloy. The rectangular apertures were created by closing the upper and lower leaf pair (see figure 3 of paper IV) and opening the middle leaf pair. The CAX depth dose was measured in a water phantom by scanning a Hi-pSi diode [101] for all apertures and CSDs described for the simulations, as well as for an open uncollimated field. Two energies were used in the experiments, 180 MeV and 98 MeV.

5.2.3 Results
In total, 18 different collimated fields with varying aperture, incident energy and CSD were simulated. Experimental measurements were performed for 14 of these fields. The simulations based on the parameterisation was compared to the corresponding full simulation by means of the gamma index, see Table 1.

In order to isolate the contribution of the collimator scattered protons in the comparison between simulations and experiments, the quotient between the CAX depth dose for the collimated and open field was compared, see Figure 16.
Figure 16. The CAX output factor, i.e. the quotient between the CAX dose of the collimated field and the open field, for the rectangular apertures. The standard deviation of the simulations is indicated by the dotted lines. The left and right column represent $E_{in} = 98$ MeV and 180 MeV, respectively.
6 Conclusions and outlook

The beam modelling methods described in this thesis show the possibility of characterising a scanned proton beam line using standard measurement equipment, such as diodes, ionisation chambers, 3D-servos and CCD-camera/fluorescent screen set-ups (alternatively film). The strength of this beam model lies in the aim of generality and the relatively compact representation, where ten parameters describe the basic properties of the elementary beam, i.e. the energy spectrum, the radial and angular distribution and the direction. Furthermore, the basic properties of the beam are described by common functions, easy to implement and to fit into the characterisation and optimisation loop of Figure 8. Also, optional and variable beam modifying elements, i.e. the range shifter and the MLC, are described by dedicated MC algorithms, which is a flexible solution that by extending the geometry descriptions of these algorithms enables the application to such elements of arbitrary shape.

The study and parameterisation of collimator scatter shows the possibility of condensing the relatively complex problem into a compact algorithm possible to handle and resulting in acceptable accuracy.

Dose distribution calculations based on this beam model show good agreement to experimental data. The comparison of dose distributions of elementary beams, Figure 9, verifies the basic properties of the beam model. The comparison of the CAX depth dose contribution of collimator scatter, Figure 16, shows good agreement for the complete beam model including all main beam modifying elements, i.e. scanning magnets, range shifter and MLC.

A prerequisite for the work presented in this thesis was the availability of experimental equipment where the scanned proton beam at the TSL, partly developed within this thesis, has been essential. However, in order to better understand complex phenomena that cannot be studied clearly by experimental means, simulations are a perfect tool, provided adequate knowledge in setting up the simulation. With this tool, problems can be dissected into its constituents and studied in detail. Both experimental and simulation aspects of proton radiotherapy have been explored within this thesis.

Scanned beam technology is still under development, and awaits the broad clinical breakthrough. In spite of the clinical expansion of proton therapy, still only one facility performs treatment with a scanned proton
beam [117]. However, almost all newly built and planned proton therapy centres have or will have at least one scanning beam line. In order to accommodate for the coming expansion, commissioning and characterisation methods such as the ones described in this thesis, need to be streamlined and adapted to fit in a clinical reality. There is still work left to be done in beam control and beam shaping for a scanned proton beams in order to take full advantage of all degrees of freedom that are offered.
Strålmodellering i dosplaneringssyfte för svepta protonstrålar

Ca 30% av alla som drabbas av någon form av cancer ordineras strålbehandling. Strålbehandling föregås alltid av dosplanering som baseras främst på CT-bilder. I dessa märks tumörvolym och strålkänsliga organ, som man i så stor utsträckning som möjligt vill undvika att bestråla, ut. Ett mål med dosplaneringen är att maximera stråldosen till målvolymen (tumören) medan man håller stråldosen så låg som möjligt i övrig, frisk vävnad. I det här avseendet har protonstrålar ett inbyggt övertag över konventionell strålbehandling eftersom protoner har en väldefinierad räckvidd och dessutom gör mest skada i slutet av sin räckvidd, i den s.k. Bragg-toppen. Dessa egenskaper gör att stråldosen till frisk vävnad som ligger i strålens ingångskanal kan hållas låg och övrig frisk vävnad kan i stort sett helt besparas från direkt bestrålning medan tumören får hög dos.

Under dosplaneringen beräknas stråldosen till patienten utifrån de fältegenskaper (bestrålningsriktningar, ordinerad dos, mm) som har satts upp. Korrektheten i denna dosberäkning beror, i stort, på tre olika processer, dels beräkningsalgoritmen i sig, hur väl den beskriver de växelverkningar mellan stråle och material som leder till deponering av dos. Dels beror det på översättningen från data i CT-bilderna till materialdata. Det tredje beroendet är beskrivningen av behandlingsmaskinens stråle, i form av positon, riktning och energi hos partiklarna, d.v.s. deras fasrymd. Denna beskrivning står strålmodellen för. Syftena och målen med den här avhandlingen var att:

- utveckla kontroll- och verifikationsmetoder för en svept protonstråle (delarbete I)
- utveckla metoder för karakterisering av svepta protonstrålar baserat på mätningar, dessa metoder ska vara generella och applicerbara på olika strållinjer (delarbete II)
- bygga upp ett ramverk för att generera fasrymden, dvs position, riktning och energi hos de individuella protonerna, för en svept protonstråle i syfte att användas som indata till en dosberäkning (delarbete II)
• studera fördelningen av protoner som sprids ut från en kollimator i syfte att parameterisera densamma så att dessa komponenter kan inkluderas i dosberäkningen i patienten (delarbete III + IV)


Jämförelser mellan dosberäkningar som baseras på den av strålmodellen återskapade fasrymden stämmer väl överens med mätningar. Detta gäller mätningar som spänner från basala mätningar av strålns grundegenskaper till mer komplexa mätningar för kollimerade svepta fält.
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A doctoral dissertation from the Faculty of Medicine, Uppsala University, is usually a summary of a number of papers. A few copies of the complete dissertation are kept at major Swedish research libraries, while the summary alone is distributed internationally through the series Digital Comprehensive Summaries of Uppsala Dissertations from the Faculty of Medicine. (Prior to January, 2005, the series was published under the title “Comprehensive Summaries of Uppsala Dissertations from the Faculty of Medicine”.)