Dose-volume effects in rat spinal cord irradiated with protons

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Chapter 4

The dose distribution

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4.1 Introduction

Since collimator scatter has a large influence on the shape of the dose distribution in small proton fields, it should be understood in detail to be able to control (e.g. minimize) the penumbra. This is in contrast to photon-therapy beams, for which collimator scatter generally accounts for less than 1% of the primary fluence [74]. A lot of work has been done to investigate similar effects in electron beams [75, 76, 77, 78]. However, the magnitude of the influence of the scatter processes for electrons differs from that for protons due to the large difference in mass.

Another effect of interest is the energy loss of the scattered protons, since low-energy protons are known to have an RBE larger than 1 [31]. This chapter reports on a Monte Carlo simulation study of the radiobiology beam line to investigate the formation of the penumbra and the effects of collimator scatter. Simulation results have been compared to profiles through 2D measurements of the dose distribution on planes downstream of the collimator.

Usually dosimetry with high spatial resolution [79] is performed with diodes, small ionization chambers [80], film [81, 82, 83], diamond detectors [84] and recently a miniature scintillating detector [85, 86]. Disadvantages of these systems (except film) are the long measuring time for obtaining a complete 2D distribution and in the case of film the time between measurement and read-out and its non-linear dose-signal relation.

In this work a CCD (charge-coupled device) camera and a scintillating screen have been used. Since spatial resolution is essential for measuring small fields, a study of the spatial response of this system has been performed. In the work of Boon et al. [72] it has been shown that the influence of the spatial response of the system can be neglected for typical clinical field sizes of 3 cm and larger. For smaller field applications, more accurate information on the dependence of
the dose to monitor-unit relation on the collimator opening was needed. Therefore, dedicated measurements of the spatial response have been performed by comparing profiles of small fields obtained with the CCD/scintillator system to those measured with a diamond detector.

4.2 Fields

The measurements of the spatial resolution were made at the isocenter, using a patient collimator of $2 \times 100 \text{ mm}^2$. For the collimator-scatter study the apertures used were in the range from $2 \times 100 \text{ mm}^2$ to $20 \times 100 \text{ mm}^2$. The slit length of 100 mm was chosen to minimize the effects of alignment errors and to allow effectively a one-dimensional analysis of the dose profiles perpendicular to the slit. These measurements were performed with the CCD/scintillator system, with screen positions at 2.5, 7.5 and 15 cm downstream of the patient collimator.

4.3 The CCD-scintillator system

4.3.1 The setup

In the spinal-cord experiments at the KVI a system consisting of a scintillating screen and a CCD camera is used [72]. The scintillating screen (Lanex fine, Kodak, Gd$_2$O$_2$:Tb) converts dose to light and the CCD camera provides a 2D image of the light pattern. The camera is equipped with a 100 mm lens that yields a resolution of 0.175 mm at the screen per camera pixel. The screen was mounted perpendicular to the beam direction and attached to the backside of a 0.5 mm thick sheet of polystyrene. The total distance between the screen and the camera via the mirror was 220 cm.

4.3.2 Image processing

The images taken with the CCD camera need some processing before the dose profiles can be extracted. Offsets are compensated by subtraction of an image taken without beam.

Isolated pixels with a large signal, due to direct interactions of neutrons and gamma rays with the CCD, are corrected by filtering the image with a median filter [87]. This filter replaces a pixel value by the median value of the $3 \times 3$ pixels environment. It is important to note that this filtering process does not influence the important features of the image, such as amplitudes and gradients [88].

4.3.3 Spatial response

In general, the presence of a non-negligible spatial response causes a “blurring” of the signal over an area of the image, which results in a decrease of the signal in the center of the field and an increase of the penumbra. For fields down to
a diameter of 30 mm, the study of Boon et al. [72] showed that the spatial response of the CCD/scintillator system did not decrease the measurement of $D_{max}$ by more than the measurement accuracy of 1% per pixel. As a worst-case estimate this would correspond to a Gaussian spatial response function with a full width at half maximum (FWHM) of 5.5 mm, in 1D. In the case of a 2×10 mm$^2$ field this would cause a decrease in output by 66%. This illustrates that more accurate data on the spatial response is needed for use in the required field-size range. To measure the spatial response of the CCD/scintillator system, the dose profile measured with the CCD/scintillator system is compared with diamond-detector measurements. We have chosen to use a diamond detector because of its high output and good linearity as compared to a diode. Note that a spatial response measured like this, includes all blurring effects in the setup, such as light scatter in the scintillator, distortions in the lens and blurring due to possible imperfect surface conditions of the mirror.

The diamond detector (PTW 60003) has a sensitive volume of a few cubic millimeters. Its spatial resolution is optimal when it is irradiated perpendicular to the axis of the cylinder, containing the sensitive volume. In that direction, the detector size is specified to be between 0.1 and 0.4 mm. An x-ray image [28] indicated that the thickness of the sensitive volume of our diamond detector is at most 0.3 mm.

Due to the large length of the slit perpendicular to the scan direction, the spatial response can be studied in one dimension. Note that for analyzing one-dimensional profiles it is sufficient to use the line spread function (LSF) of the system instead of the point spread function. First it has to be investigated whether the diamond detector behaves sufficiently like a point detector for field sizes down to the smallest field size we use (2 mm). To a good approximation the following detector LSF applies when the detector axis is perpendicular to the beam and along the scan direction $x$:

$$\varphi_{d,a}(x) = \begin{cases} 1/a & \text{for } |x| \leq a/4 \\ 0 & \text{for } |x| > a/4 \end{cases}$$

(4.1)

In equation 4.1, $a$ is the width of the sensitive volume (0.3 mm) and the subscript $d$ refers to the diamond detector. In principle, the beam profile, as measured with the diamond detector, can be deconvolved with this LSF. But since such a procedure is very sensitive to noise in the measurement, an iterative approach is used. First, an estimate of the real dose distribution is made. This estimated distribution is convolved with the LSF given by equation 4.1. The initial estimate is subsequently improved based on the difference with the measurement, until the change of the mean absolute difference of the convolved estimate with the measurement is less than 0.01% per iteration. The difference between the best estimate of the real dose distribution and the measurement indicates the influence of the spatial response of the diamond detector. In the results section it will be shown that the influence of the LSF of the diamond detector is small and can be corrected for and therefore the corrected profiles can serve as a reference for the CCD/scintillator system.
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The LSF of the CCD/scintillator system may be derived from the difference between each detector measurement for the same collimator. To model the LSF, often the sum of a broad and a narrow Gaussian is used. However, for the CCD/scintillator system under study no indication was found for a broad component [72]. Therefore, it can be assumed that the LSF of the CCD/scintillator (subscript s) system may be described by a single Gaussian distribution.

\[
\phi_{s,\sigma}(x) = \frac{1}{\sqrt{2\pi \cdot \sigma}} \cdot e^{-\frac{x^2}{2\sigma^2}}
\] (4.2)

A reference profile (deduced from the measurement with the diamond detector) is convolved with the assumed LSF. In an iterative process the parameter \(\sigma\) of the LSF is changed until the mean absolute difference of the convolved diamond profile and the screen measurement is minimized. This yields the LSF of the CCD/scintillator system, which is used to correct the CCD/scintillator profiles prior to comparison with the simulation results. The method applied to do this correction is an iterative reconstruction algorithm normally used in SPECT (single-photon emission computed tomography) reconstruction [89]

4.3.4 Spatial response measurements

A profile of a 2 mm field was measured with a diamond detector at the isocenter. Using the iterative method described in section 4.3.3 the LSF \(\phi_{d,0.3\text{mm}}\) (equation 4.1) was convolved with an estimated shape of the real dose profile.

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Figure 4.1: The dose profile of a 2 mm field, 15 cm behind the collimator as measured with the diamond detector. The profile corrected for the spatial response of the diamond detector shows a less than 1% higher dose.

A profile of a 2 mm field was measured with a diamond detector at the isocenter. Using the iterative method described in section 4.3.3 the LSF \(\phi_{d,0.3\text{mm}}\) (equation 4.1) was convolved with an estimated shape of the real dose profile.
The resulting profile is shown in figure 4.1 and it can be seen that the deviation introduced by the instrument response function is about 0.6% at maximum dose. This indicates that the spatial response of the diamond detector has only a very small influence on the measured profile and this influence can be corrected for. The corrected profile is used as the reference profile for the CCD/scintillator measurements.

The iterative method described in section 4.3.3 has been used to determine the LSF of the CCD/scintillator system. It was found to be a Gaussian distribution with $\sigma=0.22$ mm (see equation 4.2). A comparison between the reference profile and the (deconvolved) CCD/scintillator measurement is shown in figure 4.2. For the optimal LSF the mean deviation of the convolved reference profile and the CCD/scintillator measurement is only 0.75%.

In this study it is shown that the spatial response of the diamond detector has a small (0.5%), correctable influence on the measured $D_{max}$ of a 2 mm profile. To get an indication of the influence on measurements in smaller fields, dose profiles were constructed, using two error functions to describe both penumbrae. The penumbra of the larger fields was matched to the 20 mm slit measurement. These profiles were convolved with the LSF of the diamond detector. The result of the measured value of $D_{max}$ as a percentage of the real $D_{max}$ is shown in figure 4.3. Here, it can be seen how the detector effect on the measured $D_{max}$ increases for fields smaller than 2 mm. With a slit size of 1 mm its magnitude increases.

Figure 4.2: The dose profile of a 2 mm field, 15 cm behind the collimator as measured with the CCD/scintillator system. From this measurement the true dose profile is obtained by the deconvolution procedure as described in the text. For comparison the measurement with the diamond detector, corrected for its LSF, is also shown.
is about 1%.

The LSF of the CCD/scintillator system is well described by a Gaussian distribution with $\sigma=0.22$ mm. For applications in small-field dosimetry it is important to know how this influences the results as a function of the field size. In figure 4.3 the measured value of the maximum dose in a profile is shown as a function of the slit size. These values were calculated using the same method as described for the diamond detector. It can be seen that for slit sizes larger than 2 mm the deviation is less than 2% and for field sizes larger than 2.5 mm the deviation becomes negligible.

4.4 Monte Carlo simulations

4.4.1 The Monte Carlo code

In a Monte Carlo simulation all types of scatter contributions can be registered separately. Collimator scatter can be divided into several categories (figure 4.4) that define the particle trajectories to and from the collimator. Scattered particles in each category have a different effect on the dose distribution downstream of the collimator. Part of the simulation was carried out using the Monte Carlo code library GEANT, version 3.21. This code has been developed for nuclear and high-energy physics applications and it has been shown that it can also be applied successfully in the energy range used for proton therapy [90, 91].

The simulated setup is depicted in figure 4.4. Protons start their trajectory

![Figure 4.3: Influence of the spatial response on the $D_{max}$ as a function of the field size as would be measured by a 0.3 mm thick diamond detector (dots) and the CCD/scintillator (line)
at the entrance face of the first scatter foil. Different diameters of the simulated proton source were used to investigate the influence of the initial beam size and it was found to be negligible. Also, the initial angular spread of the beam was set to zero, since it was very small compared to the angular spread introduced by the scattering process. The beam interaction with the dual-scattering system was simulated in 3D with a separate code that is based on Highland’s approximation [92, 93] of multiple scattering theory [94], labeled foil code. In this condensed-history approach, the actual scatter angles of the protons were calculated using a random generator with a normal distribution. The random generator is based on the random generator GRNDM in the GEANT library. The known starting positions and the first scattering angles determine the incident spatial positions of the protons on the second scatter foil. The second foil has an inhomogeneous shape, designed to obtain a homogeneous dose distribution at the isocenter. Therefore, the second scattering process is sensitive to the impact position on this foil.

The output of the foil code is interfaced as input to GEANT for further
transportation of the protons through air towards the patient collimator. Both
the number of protons that arrive downstream of the patient collimator and
their energy loss are registered in a 1.0 mm thick layer of water to determine
their contribution to the dose. We thus calculate the dose as measured with a
water-equivalent detector of 1 mm thickness, placed in air.

Three distinct categories are sampled. The first category is for protons that
did not have any interaction with the collimator (region 1 in figure 4.4.) The
second category is for protons that hit the collimator, but scatter out of the
material through the collimator inner face to reach the plane of measurement.
The protons from this second category are sub-categorized into 2a: those that
are initially incident upon the collimator inner face and 2b: those that are
incident upon the collimator face at the source side (entrance face). The third
category consists of those protons that are incident on the collimator source-
side face at more than 4 mm from the edge of the aperture (figure 4.4). The
simulations show that a proton in the third category never reaches the plane
of measurement and so it is discarded without further simulation or registration.
Note that the patient collimator is the only collimator in the simulation. All
other beam-limiting devices are designed for a beam of 8 cm diameter, their
influence on a small field is considered negligible.

Simulations were performed for different collimator-to-detector distances.
The influence of the air in the setup on the penumbra has been investigated by
comparison to simulations with a setup in vacuum. To investigate the influence
of collimator scatter on the radiobiological effectiveness (RBE) and its spatial
distribution, the energy of the protons after interaction with the collimator was
registered.

A typical simulation used $10^7$ proton histories, of which about 10%-20%
reached region 1 or 2. It required 7 hours on a DEC 500 MHz Alpha station.

4.4.2 Comparing the simulations with measurements

To validate the Monte Carlo simulation of the scatter process, the simulations
are compared to measurements. In figure 4.5, examples of this comparison
are shown for several fields at several distances from the collimator. In figure
4.5a, profiles at the collimator exit are shown. In this figure it can be seen
that the correspondence of the simulations to the measurements is excellent,
apart from a slight deviation in the penumbra. This can be explained by the
extreme sensitivity of the penumbra to the actual distance to the collimator at
the collimator exit and the spatial resolution of the detector. The latter effect
is corrected for, but some small artifacts may have remained [95].

In figure 4.5c, dose profiles at 15 cm from the collimator are shown. The
agreement of the simulations with the measurements is good. This includes the
steepness of the penumbra. The maximum error is found in the tails of the 20
mm field. In figure 4.5b, the intermediate distributions at 7.5 cm behind the
collimator are shown, for the 20 mm and 2 mm fields, which also show good
agreement.

From the data presented in figure 4.5 it was concluded that the Monte Carlo
Figure 4.5: Comparison of measurements (CCD/screen, dots) and simulations (lines) for several field sizes and distances to the collimator. Note that for clarity of display the 4 mm, 8 mm and 20 mm profiles were given an offset of 0.2, 0.4 and 0.6, respectively.
simulations are in agreement with the measurements within 2.5%. A very sensitive property of the simulated dose distributions is the penumbra. In the same figure it can already be seen that a very good agreement is found. Also, the significant (up to 7%) local-dose increase outside the 2 mm field agrees with the measurements both in level and in structure within the noise of the simulation. It may thus be expected that also other geometries can be simulated with similar accuracy.

4.4.3 Scatter contributions

Figure 4.6: A cumulative plot of the dose components in a simulated 20 mm field at the exit of the collimator.

In the simulation the different scatter contributions have been stored. The dose distribution immediately at the exit of a 20 mm slit is depicted in figure 4.6. Here, it can be seen that protons entering the collimator through entrance face and leaving it from the inner face have a rather large contribution to the dose distribution: the contribution of the scatter ears or horns is 20% of the total dose. The scatter on the inner face of the slit has a contribution of comparable magnitude (15%) only in a small region in the direct proximity of the collimator edge. A close inspection of the spatial distribution shows that this contribution is also present behind the collimator material, although not more than 3 mm from the aperture.

In figure 4.7, the dose distribution 15 cm downstream of the slit is shown. Here, it can be seen that the inhomogeneity caused by the entrance-face scatter has disappeared. The entrance-face scatter is spread over the entire field, adding about 5% to the dose of the unperturbed beam. The spread in the inner-
face scatter is much less. It is still concentrated close to the projection of the collimator wall and contributes locally about 2.5-3% to the total dose.

Figure 4.7: A cumulative plot of the dose components in a simulated 20 mm field at 15 cm behind the collimator.

Figure 4.8: Relative dose on the beam axis as a function of the distance from the collimator in a 4 mm field and a 20 mm field.
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The contribution of entrance-face scatter to the dose distribution changes with distance from the collimator. In figure 4.8, a plot is shown of the dose on the central axis as a function of the distance to the collimator. In figure 4.9, a complete overview of the evolution of the different contributions to the dose distribution is shown. Note that the gray scales are normalized per contribution. In the undisturbed part of the beam it can be seen that the penumbra increases gradually when moving away from the collimator. Also, the maximum dose decreases due to the divergence from the virtual source. The entrance-face scatter contribution is well localized, but spreads over the dose distribution very fast. The inner-face scatter is very localized and shows little spread.

4.4.4 Air scatter

By default the simulations were carried out with the beam travelling through air. To investigate the influence of air on the penumbra, simulations of the beam line in vacuum were also performed.

The variation of the penumbra (20%-80%) as a function of the distance to the collimator is shown in figure 4.10. In air the penumbra increases with 0.054 mm per cm distance from the collimator. In vacuum the penumbra increases with 0.013 mm per cm. The offset of about 0.12 mm is caused by the fact that the bin-size in the simulations was 0.2 mm.

In the same plot the penumbra of the total dose distribution is also plotted for air. It can be seen that the maximum difference with the penumbra of the direct beam is only 0.08 mm. Note that the variation in difference is fully explained by the bin size and noise in the simulations.

From the change in lateral position of the 50% dose level, the position of the virtual source can be derived [75]. The virtual source is at the position where the extrapolation of the 50% dose crosses the beam axis. In vacuum we find the source-to-collimator-exit distance to be 3.2 m, and in air it is 3.0 m. Back projection of the 20% and 80% dose levels gives an estimate of the virtual source size. For the simulations in vacuum the effective source size is 4 mm. In

![Figure 4.9: Evolution of the different components of the dose distribution of a 20 mm field as a function of the distance from the collimator. Note that each plot is normalized to its own maximum.](image)
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According to Highlands formula the standard deviation of the Gaussian-shaped scatter cone due to 3 meters of air is 4 mrad. Therefore, air scatter between source and collimator has a contribution, which is at least as large as the contribution of the source size to the penumbra. This implies that insertion of vacuum sections to the beam line will improve the penumbra. Alternatively one may use helium-filled beam-line sections, similar to that already applied in the Scanditronix MM50 racetrack microtron [96, 97].

4.4.5 Position vs. energy distribution

In a simulation of a 20 mm field 2.5 cm behind the collimator the energy of the scattered protons was registered. In figure 4.11, the solid and dotted lines show the energy spectra for the entrance-face and inner-face scattered protons, respectively. Both are expressed in particle fluence per energy bin.

From this figure it can be seen that the energy of the protons emerging from the inner-face scatter remains larger than 40 MeV for 96% of the protons. For the entrance-face scatter the distribution is much broader. The mean energy after entrance-face scattering is about 50% of the beam energy.

To investigate the contribution of the different proton energies to the dose, the particle fluence curves were multiplied with the energy-dependent stopping power of protons in water. The resulting curves were normalized to the maximum of the sum of both scatter contributions and plotted with a dash-dotted line (inner-face scatter) and a dashed line (entrance-face scatter). The plots

Figure 4.10: Evolution of the 20% to 80% penumbra as a function of the distance to the collimator. The undisturbed beam consists of particles that are in region 1 in figure 4.4.
show that, although the number of low-energy protons is small, they contribute most to the dose.

Figure 4.11: Energy spectra expressed in relative particle fluence per energy bin, integrated over a 20 mm field resolved for the different (entrance- and inner-face) scatter contributions (solid line and dotted line) at 2.5 cm behind the collimator. The relative dose deposited for each particle energy is also plotted (dashed line and dash-dotted line). Both the energy spectrum and the dose curves are normalized to the maximum of the sum of the absolute curves. Thus, the relative contributions of the inner-face and entrance-face scatter curves can directly be read in the plot.

Figure 4.12: Spectral maps of the 20 mm field at 1.25 cm behind the collimator for both scatter contributions. Note that the gray-scale is normalized separately for each plot.
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In figure 4.12 the positional dependence of the energy spectrum is shown for both scatter contributions. Here, it can be seen that the majority of the protons in the scatter ears have energies above 40 MeV. Also, it can be seen that the protons with an energy lower than 40 MeV are spread rather homogeneously over the dose distribution.

4.4.6 Analysis of collimator scatter

![Diagram showing dose profile](image)

Figure 4.13: Simulated dose profile 1.25 cm downstream of a 1-degree-tilted collimator.

One of the aims of this Monte Carlo study was to determine how the collimator scatter influenced the penumbra and the field shape in general. From the simulations it was found that the penumbra of the undisturbed beam mostly determines the penumbra of the total dose distribution. Therefore, we conclude that in our setup the penumbra is totally determined by geometrical and air-scatter effects of the primary beam. This means that optimization of the penumbra is achieved by simply staying close to the collimator. However, when coming closer to the collimator the dose distribution becomes inhomogeneous and the RBE may vary across the field. Therefore, the collimator-scatter contributions need to be taken into account.

If one looks at the two scatter contributions in a 20 mm field (figures 4.6 and 4.7), the main contribution is the entrance-face scatter. In general, it can be described as a separate beam which contributes about 20% to the total dose at the collimator exit and it is aimed at an angle of 6 degrees towards the center of the direct beam (figure 4.9). When moving further away from the collimator, this scattered beam spreads fast and its contribution to the dose decreases. However, it is important to note that the dose on the beam axis first
increases to a maximum at 2.5 cm (figure 4.8), and decreases slowly at larger distances from the collimator. In a larger field (e.g. 20 mm) the dose increases until the scatter ears have merged. In smaller fields the peaks coincide already at the collimator exit and the dose on the central axis will only decrease for increasing distance (see figure 4.5).

Note that in a scatter ear the relative dose increases due to the scattered particles and is, in principle, independent of the aperture-size. At any distance it contributes with an aperture-size-independent flux to the direct beam, that has a fixed flux as well, because the amount of irradiated collimator material that contributes to the dose remains the same (in 1D).

The relative contributions of the different types of collimator scatter are dependent on setup parameters. For example, in a setup where the virtual source is closer to the patient collimator the importance of inner-face scatter will become more important.

The thickness of the collimator is only of influence on the inner-face scatter contribution, as it increases the solid angle of the inner face of the collimator, but not the escape probability of the protons entering the entrance face of the collimator. Also, collimator alignment has some influence on the field shape. The inner-face scatter is proportional to the number of protons that hit the inside of the collimator. Misalignment of the collimator increases the solid angle covered by one collimator edge and decreases the solid angle of the opposing edge, thus introducing an asymmetric contribution of the inner-face scatter. The entrance-face scatter is dependent on the probability that a proton escapes from the edge after entering from the entrance face. This probability depends on the range of multiple scatter angles that result in escape. A tilt of the collimator decreases the size of the angular range in which a proton would escape, while in the opposing edge, the range gets wider.

To show the influence of misalignment on both scatter contributions a 40 mm field was used, in order to separate the scatter contributions on both edges of the field. In figure 4.13 a profile at 1.25 cm behind a 1-degree-tilted collimator is shown. The tilt reduces the entrance-face scatter and increases the inner-face scatter at the right side of the field. The dose tail behind the collimator at the right side of the field is characteristic for misalignment, since inner-face scattering is increased a lot at one side. This was not observed in the measurements. Therefore, it is concluded that during the measurements the alignment of the collimator was better than 1 degree.

4.4.7 Energy and RBE considerations

If we use the model as employed by Paganetti et al. [31], it can be assumed that the RBE is 1 for protons over 40 MeV. The RBE increases more or less linearly to 1.2 as the energy decreases towards zero. The absolute RBE values are, of course, depending on the biological endpoint. From figure 4.11 it can be seen that about 40% of the scatter dose is deposited by protons with RBE>1. At 1.25 cm from the collimator the dose contribution from collimator scatter is about 13%. From figure 4.12 it can be derived that 40% of the dose of these
protons comes from protons with increased RBE. However, it is also found that these low-energy protons are spread homogeneously over the field, so that the RBE effects are diluted. At a distance of 1.25 cm from the collimator only about 5% of the dose is deposited by protons with RBE > 1. If we adopt the assumption that the maximum RBE for protons of more than 5 MeV is 1.2, the biological damage to tissue is estimated to be only 1% larger than expected from the physical dose.

4.4.8 Non-linear response of Lanex to low energy protons

It is well known that scintillators show non-linear behavior as a function of stopping power. In a study of collimator scatter it is therefore important to have a closer look at the response of the screen to different proton energies to get an impression of the accuracy of the CCD/scintillator system for collimator-scatter measurements. For a worst-case estimate of the deviation between the CCD/scintillator measurement and the dose distribution the profile of a 20 mm field at the collimator exit was used.

The energy spectra presented in figure 4.12 were used to determine the stopping-power distribution. Subsequently, quenching was applied to the distributions using Birks formula [99, 72]. The quenched profile was normalized to the dose profile. The maximum difference between those profiles was found outside the field. The difference was 10% of the local dose. There the dose is well below 10% of the central dose in the field, and therefore this effect may be neglected as the difference is at most 1% of the dose in the field.

Inside the field the maximum deviation in the scatter contribution was 5%. This deviation was found on the beam axis. There the contribution of scatter to the total dose is about 10%. This means that the maximum deviation between a CCD/scintillator measurement and a dose profile is at most 0.5%-1% of the maximum dose, inside and outside the field, respectively.

4.5 Conclusions

The LSF of the scintillating screen and the CCD camera can be described by a Gaussian with $\sigma = 0.22$ mm. This spatial response enables direct use at field sizes larger than 4 mm whereas profiles in fields down to 2 mm can be measured with a small (<3%) correction. This makes the scintillating screen a very suitable instrument for dosimetry in small fields and it also allows accurate validation measurements of Monte Carlo simulations.

The GEANT simulations show a very good agreement with the measurements with the CCD/scintillator system. From the simulation it follows that in our scatter-foil/collimator system, the entrance-face scatter is the most important scatter contribution to the dose distribution, but it is important to note that it has no significant influence on the penumbra. The influence of the collimator scatter on the RBE is at most 1% at 1.25 cm behind the collimator,
although the dose inhomogeneity from the entrance-face scatter is about 20% there.

The influence of air scatter in the beam line is the major contributor to the penumbra. Compared to a geometrical (vacuum) penumbra of 0.3 mm at the isocenter, the presence of air causes the penumbra to increase to 1.1 mm. The combination of collimator scatter, air scatter and geometry results in a trade-off between penumbra sharpness and field homogeneity. In our setup a good compromise occurs at a distance of 4-5 cm behind the collimator, where the 20%-80% penumbra is 0.5 mm.

From the above results and comparison to the data we conclude that Monte Carlo simulation is a useful and reliable tool to investigate different properties of small irradiation fields in detail.