Dosimetry and quality control of scanning proton beams

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5

Applications of the CCD + scintillator system

5.1 Introduction

In this chapter we will discuss two applications of the CCD system described in chapter 4. One deals with the special problems involved in the quality control of a spot scanning beam delivery system, the other with the verification of a treatment plan that contains an inhomogeneity. The measurements described in this chapter have been performed mainly at the PSI spot-scanning gantry in Switzerland [103], which will be described in brief in section 5.2.1. It should be noted here that the PSI spot scanning control system contains already a sufficient amount of other monitoring devices to prevent errors in the dose delivery, but the purpose of this study is to show that the CCD system is able to perform as an extra, independent quality control device.

After we have verified in section 5.3 that the CCD + scintillator system properly functions as a dosimeter in a spot-scanning beam, we will investigate in section 5.4 the QC capabilities of the CCD + scintillator system. We start there with a study of the spot positioning in a plane. The relation between dose homogeneity, spot size and inter-spot distance is investigated. Subsequently the study is extended to a volume by adding proton beams with different energies. We will show to which extent missing or double spots can be detected. It will be shown that although the system is essentially two dimensional, the dose at the screen position is correlated to the dose deposition at larger depths. Even if a scan error occurs for a pencil beam whose Bragg peak (i.e. the most important contribution to the total dose) lays behind the screen, it is still possible to detect this error from the total dose pattern observed at the screen position.

Because of the high spatial resolution and large sensitivity of the CCD system, interesting effects can be observed which are caused by changes in the proton flux due to differences in multiple scattering effects (see also section 2.1.4) for different materials. An example of this effect is discussed in section 5.5: the verification of a treatment plan of the dose distribution in an inhomogeneous region, which is difficult to verify with conventional dosimetry equipment.
5.2 Experimental setup

5.2.1 The spot-scan technique

PSI is the second place in the world where a proton gantry is installed. With its 4 m diameter it is much smaller than the gantries at Loma Linda (12 m diameter). The compact gantry became possible by using some technical novelties which are described in detail in [103, 113, 115, 116]. For the work reported in this chapter the most important feature of the PSI gantry is the active beam steering system, which uses the spot-scanning technique. The main elements used in this system are schematically shown in figure 5.1.

The dose distribution in the isocentre of the gantry, that is the position of the target volume (which can either be a tumor or a phantom used for dosimetric measurements) is determined by the sum of individual spots which have a certain weight and a certain

![Figure 5.1: Schematic overview of the PSI gantry. Scanning along the U axis is performed by scanning magnet, along T axis (⊥ paper) by table and along S axis using range shifter plates. More description of the Cartesian S,T,U coordinate system in the text.](image-url)
position. The spot position is given in a Cartesian coordinate system \( S, T, U \) which is described below and the weight of a spot is determined by the integrated beam current. The word spot may be somewhat misleading, since it suggests that the dose is confined to a volume (i.e. the Bragg peak), which is obviously not the case. Spots are extended along the \( S \)-axis from the entrance to the position of the Bragg peak (perhaps the word pin would be more appropriate). The \( S \)-coordinate describes the position of this maximum. The \( T \) and \( U \) coordinate are the lateral position of the centre of the spot relative to the isocentre. The steps involved in the delivery of one spot are:

1. a fast kicker magnet is able to switch the beam on and off in less than 50 \( \mu s \). This determines the number of protons that is delivered in one spot.

2. a sweeper magnet bends the beam in front of the 90° bending magnet in the gantry head. Due to the ion-optical properties of the bending magnet this causes a parallel shift of the beam at the position of the isocentre over \( \pm 10 \) cm. The time needed to perform a change of the sweeper settings is 3 ms. According to PSI convention this direction is defined as the \( U \)-axis.

3. after the 90° bending magnet the beam leaves the vacuum and is slowed down by a variable number of range shifter plates, which can be moved into the beam using a pneumatic valve system. There are 36 range shifter plates each with a thickness of 5 mm polyethylene (water equivalent thickness: 4.5 mm) and one 2.5 mm thick plate. In this way the Bragg peak can be shifted in water from 0 to 16 cm depth with a resolution of 2.3 mm. It takes 30 ms to change the range shifter plate setting. This direction is defined as the \( S \)-axis.

4. the positioning of the spot in the other lateral direction, perpendicular to the magnetic scan direction, is performed by moving the patient-table. This direction is defined as the \( T \)-axis. In principle this could also be done with a second sweeping magnet. The reason for not using such a double scanning magnet system is that this would require a much bigger 90° bending magnet, which means that the gantry would be much less compact. The time involved in changing the table settings is the longest of all, 1-2 s. The accuracy of positioning is better than 0.1 mm.

The steering of the spots is performed using an advanced, fully computerized control system. The order of scanning is chosen in such a way that it minimizes the treatment time: changes along the \( U \)-axis occur most frequently, and changes along the \( T \)-axis the least.

The time scales involved in the scanning system are illustrated by the following ‘reference scan’. A homogeneous \( 6 \times 6 \times 6 \) cm cube is irradiated with 2 Gy using spot depositions applied at 5 mm intervals, giving a total of \( 15^3=3.4 \times 10^3 \) spots. The time per spot varies from 5 ms to 100 ms, depending on the number of protons that have to be
applied (the beam current is \( \approx 0.1 - 0.2 \text{ nA} \)). The complete time necessary to scan this box is 3.5 minute. The actual time during which dose is delivered is however shorter, since the 3 spot-positioning systems have a dead time. The duty cycle of the system is determined by the dead time of the 3 systems, and by their relative occurrence. For the reference scan the duty factor will be \( \approx 60\% \), which means that only during 2 minutes there is actual beam delivery.

5.2.2 Spot scan monitoring

It is clear that the spot-scanning technique is a very flexible system, which allows very good conformation of the dose distribution to the target volume. Because of its complexity, however, it requires also sophisticated, on-line beam monitoring instruments in order to ensure a safe treatment. In the PSI gantry the following systems are present:

- two independent beam intensity monitors which count the number of protons. These are transmission ionization chambers positioned before the range shifter plates. Because of the small irradiation times in the spot scan system (down to \( \approx 5 \text{ ms per spot} \)) the system has to be very fast. This can be realized by using an ionization chamber with a small airgap, but in that case the signal is also small. It was therefore chosen to have one ionization chamber with a small gap (5 mm) which determines when a spot is completed. The ion collection time of this chamber is less than 100 \( \mu s \). Another monitor, consisting of an ionization chamber with a broader gap (10 mm) provides a more accurate measurement of the integrated beam current of a spot. This output is used for quality control purposes. Moreover it provides an additional interlock, in case the first monitor would fail.

- a position sensitive monitor that covers the full swept beam. It is a transmission ionization chamber of which the signal plane is divided in 48 aluminium strips which have a width of 4 mm. The monitor causes an interlock in case the readout position of the beam does not correspond with the planned position.

- a Hall probe in the sweeper magnet monitors the magnetic field and thus the beam position.

- the functioning of the range shifter plates is verified using an optical system.

- an independent system of ‘watch dogs’ is installed that ensures that a certain maximum dose per spot (equivalent to 100 ms beam time) will never be exceeded.

An interlock in one of these subsystems will immediately causes the kicker magnet to stop the beam (within 50 \( \mu s \)). For additional safety also a mechanical beamstop will
be closed, which system of course has a longer reaction time (50 ms). The safety system described above has been thoroughly tested and it performs well.

According to the quality assurance and quality control standards (see section 1.2) it is desired to have as much independent verification as reasonably can be achieved. A system that can perform an easy, day-to-day quality control will therefore be of help. We will describe in section 5.4 to which extent our CCD-scintillator system is suitable for this.

5.2.3 CCD system setup

The CCD-scintillator system (as described in section 4.3.1 and shown in figure 4.1) was positioned in its entirety on the patient table. The gantry angle was turned by 90° in order to irradiate the phantom from the side. The movement in the T direction was performed by moving the table up and down (with the CCD system on top of it). The distance between the last range shifter plate and phantom/screen combination was 40 cm. For the variation of phantom depth we have used two methods:

- inserting a variable amount of polystyrene slabs before the scintillator screen, this is the method we have also used for the experiments in chapter 4.

- using a water bellows system, designed and build at PSI, to which the screen is attached. In this system a variable amount of water is pumped into a bellows system, which has thin PMMA (=lucite) windows. In this way it is possible to have a variable amount of water in front of the scintillator screen, which remains at a fixed position.

In both cases no phantom material behind the scintillator has to be used, since the effect of backscatter is small for protons (see section 3.5.4).

The water bellows system turned out to be a very useful addition of the CCD system, since it provides remotely controlled variation of depth with a resolution of 0.5 mm. The complete measurement can be performed in this way without interruptions (unlike the situation where polystyrene slabs are used, in which case the treatment room has to be entered after each depth in order to add/remove a slab). In order to make the CCD system suitable for routine use, a bellows system should be added to the system.

Thanks to the low-dark current of the CCD camera (see section 4.3.1) it was possible to acquire a complete scan (which lasts up to 5 minutes) in a single CCD measurement.

The measurement of the reference dose was performed using two instruments:

- a calibrated Markus (plane-parallel) ionization chamber mounted to the beam exit side of the screen.
a large area (\(\varnothing=80\) mm) ionization chamber, attached to the water bellows system. With this chamber also the measurements described in section 3.3 have been performed. This chamber is very suitable for measuring depth-dose distributions, since it gives a large signal.

5.3 Dosimetric characteristics

The light yield measurements as a function of applied dose at passively scattered beams show reasonable agreement with calculations (see chapter 4). The dependence on dose rate has not been investigated in that case, since the dose rate in passively scattered beams is more or less uniform and does not exceed 1-2 Gy/min. In a spot scanning system, however, the dose rate at the centre of a spot can be as high as 300 Gy/min. One has to verify therefore if the light output per unit dose remains constant. In the first part of this section we compare the ratio measured/calculated yield in the low ionization density region (1 cm depth) for both the PSI and the Uppsala beam, with a high and low dose rate respectively. In the second part we will investigate the effect of higher ionization densities at the same high dose rate, by measuring the dose as a function of depth with the CCD system and with an ionization chamber. In this way the decrease of light yield in the Bragg peak can be determined.

5.3.1 Dose-rate effects

To estimate if a dose rate effect can be expected, we will calculate the probability that two protons pass within the range of a doping centre (2 nm according to [94]) and within the occupation time of a doping centre in an inorganic scintillator (< 10 \(\mu\)s, according to [94]). From table 2.1 it can be derived that 177 MeV protons have an energy loss of \(\approx 6\) MeV cm\(^{-2}\) g\(^{-1}\) at a depth of 1 cm polystyrene. This implies that a homogeneous dose of 1 Gy at that depth corresponds to \(1.05\cdot10^9\) protons per cm\(^2\).

Using Poisson statistics, it can be calculated that the probability that 2 protons will pass the same doping centre for a dose rate of 1 Gy/s is smaller than \(10^{-9}\). This means that the dose rate has to increase by orders of magnitude in order to have effect on the light production efficiency (in which case also the radiobiological effectiveness is expected to change).

Experimentally we measured the dose rate effect by comparing the CCD yield with the reading of a calibrated Markus chamber at the position of the screen. For the screen we have used Gd\(_2\)O\(_2\)S:Tb (or Lanex, see section 4.1.3). Both the screen and the Markus chamber were positioned behind a 6 cm polystyrene phantom. The applied dose distribution was a homogeneous 10\(\times\)14 cm\(^2\) field, produced with an inter-spot distance of 5 mm and spot width of 7 mm FWHM, without using range shifter plates (more details on the production of such a field will follow in section 5.4.2). The resulting conversion factor of light yield to dose was 2.44\(\cdot\)10\(^4\) ADU/Gy. In order to enable comparison with
the passively scattered Uppsala beam, we have divided the result by the calculated
yield (using equation (4.3) and tables 4.2, 4.3, see section 4.2.1). The result is:

\[
\text{PSI measured/calculated: } 1.14 \pm 0.02 \tag{5.1}
\]

while the ratio for the passively scattered beam was:

\[
\text{Uppsala measured/calculated: } 1.12 \pm 0.02 \tag{5.2}
\]

We have repeated the same measurement 6 months later and the ratio between the
measured and calculated value was within 1.3 % consistent (see also table 4.6 in sec-
tion 4.4.1).

5.3.2 Dose linearity

Another confirmation of the linearity of our CCD system came from the measurement
of an inhomogeneous dose distribution, a \textit{wedge} (cf. figure 4.7). The result, together
with a horizontal profile, is shown figure 5.2. It is produced by varying the weight
of the spots (the number of protons per spot) linearly with the distance along the \text{T}-
axis. The linearity between the number of protons and the CCD signal is good: by
performing a fit, we obtained a gradient of $582 \pm 6 \text{ ADU/cm} \equiv 2.4 \pm 0.03 \text{ cGy/cm}$
(using the conversion factor of light yield to dose from above)\textsuperscript{1}. The quoted uncertainty
is determined by the quality of the fit. The programmed proton flux (protons/cm\textsuperscript{2}) in
the steering program varied from $4.82 \cdot 10^8$ to $1.51 \cdot 10^7$ over a distance of 20 cm, which
 corresponds to a dose gradient of 2.2 cGy/cm using the number of protons cm\textsuperscript{-2} Gy\textsuperscript{-1}:
$1.05 \cdot 10^9$. The 10 % difference between the two values can be explained by uncertainties
in the conversion factor from number of protons to dose (which makes assumptions on
the proton energy/energy loss) and uncertainties in the light conversion factor.

5.3.3 Ionization density dependence

The next step was to investigate the ionization density dependence. We have performed
this by measuring the depth-dose distribution with the CCD system and with an ion-
zation chamber. To vary the depth we have used the PSI water bellows system. For the
reference signal a large ($\beta=80$ mm) ionization chamber was used, which was attached
to the beam exit side of the bellows system. The measurement was repeated with the
screen attached to the bellows system. The results can be seen in figure 5.3. In the same
way as in section 4.3.5 the results were normalized to a dose per fluence.

The decrease of the light signal in the Bragg peak is 5 %. This is less than for
the passively scattered Uppsala beam (8 %, see section 4.4.3). Since the 177 MeV
beam we have used for this experiment is degraded from a 600 MeV beam, the initial
energy spread is considerable (according to the MC calculations in chapter 3: 0.9 %

\textsuperscript{1} 1 \text{ cGy} = 10^{-2} \text{ Gy} = 1 \text{ rad}
Figure 5.2: Wedge dose distribution measured with the CCD system at a depth of 1 cm polystyrene. The wedge is produced by varying the monitor units (number of protons) linearly with the position in the $T$ direction.

Figure 5.3: PSI depth-dose distribution in water. Continuous line: measurements with a large ionization chamber ($\Theta=80$ mm) operated at $HV=300$ kV; diamonds: CCD measurement with $\text{Gd}_2\text{O}_2\text{S}:\text{Tb}$ scintillator and dotted line: the quenching formula (4.1) with $k_B=4.0$ mg cm$^{-2}$ MeV$^{-1}$ applied to the ionization chamber data.
of the initial energy for the PSI beam vs. 0.5 % for the Uppsala beam). This causes a broadening of the Bragg peak (see section 2.1.6). The mean proton energy in the Bragg peak is also higher. Both effects contribute to a smaller decrease of the light yield (see sections 3.5 and 4.1.2). The correction needed to convert the light output to a dose for the PSI beam is therefore small, and can be done using the procedure in section 4.3.5. The resulting effective $k_B$ for the PSI beam yields $4.0 \text{ mg cm}^{-2} \text{ MeV}^{-1}$. The systematic uncertainty in the dose resulting from a measurement with the CCD system is mainly determined by the uncertainty in this $k_B$ value. When we set $k_B = 0$, which gives the maximum possible error, the maximum error of 5 % occurs at the distal edge.

5.3.4 Conclusion

We have shown that the system is able to function as dosimeter in spot scanning beams, since the ratio measured/calculated yield at the entrance agreed within 2 % with measurements in passively scattered beams, for which the dose rate is about a factor $10^{-2}$ lower. The linearity of the CCD system versus dose was measured to be better than 1 %. Also in regions with a higher ionization density ($\approx 20 \text{ MeV cm}^2 \text{ g}^{-1}$) the dose rate has no influence, since the observed light decrease in the Bragg peak is only 5 %. We have also demonstrated the stability of our system by repeating a measurement after 6 months which yielded results that agreed within 1.5 %.

5.4 Spot scan verification

In this section we will demonstrate the capabilities of the CCD system for quality control of a spot scanning system. We will start with the measurement of a single spot, continue with the dose in a plane and end with the measurement of a dose volume. The emphasis is on measurements that are performed at fixed depth, but which can detect errors that occur during the complete scan, i.e. at all depths. This enables the quick verification of the treatment plan, as described in section 5.2.2.

5.4.1 Elementary spot measurement

First, we have measured a single spot with the CCD system. This measurement was performed at a depth of 1 cm polystyrene. We used $1.21 \times 10^8$ protons for a single spot, as set by the steering program.

The measured results are shown in figure 5.4. Using the conversion factor of light yield to dose obtained in section 5.3: $2.44 \times 10^4 \text{ ADU/Gy}$, the result measured with the CCD system in the centre of the spot is 0.15 Gy. By fitting a Gaussian to the horizontal and vertical profiles it was found that $\text{FWHM}$ in the horizontal direction was $7.6 \pm 0.1 \text{ mm}$, and in the vertical direction $7.6 \pm 0.2 \text{ mm}$. The horizontal and verti-
Figure 5.4: Image of elementary spot, together with horizontal and vertical profiles. In the profiles also the result of a Gaussian fit is shown (dotted line)

cal widths are equal, but the uncertainty in the vertical fit is larger, since the profile deviates more from a Gaussian.

The elementary pencil beam is not symmetric: the profile shows a tail in the lower right corner, and the asymmetry in the U (scanning magnet) direction is worse than in the T (table) direction.

It is interesting to see whether the dose in the centre of the spot is in agreement with the applied number of protons per spot. Therefore we have to derive the relation between the dose in the centre of a single spot and the total dose in a homogeneous dose distribution, since it is not possible to directly measure the dose in the small area corresponding to the spot centre. This is easiest done by a numerical calculation, in which the homogeneous dose distribution was constructed by using Gaussian shaped spots with a FWHM of 7.6 mm and an inter-spot distance of 5 mm. The result of such a calculation is shown in figure 5.5.

The numerical calculation shows that the value in the centre of a spot is 34 % of the value in the plateau of summed Gaussians. A number of $1.21 \cdot 10^8$ protons per spot (according to the PSI treatment program) corresponds to a homogeneous proton flux of $4.82 \cdot 10^8$ cm$^{-2}$, and thus to a homogeneous dose of 0.46 Gy using the number of
protons cm\(^{-2}\) Gy\(^{-1}\): 1.05 \(\cdot\) 10\(^9\). The expected value in the centre of the spot is then 34 % of 0.46 Gy = 0.16 Gy, which is within errors in agreement with our CCD measurement of 0.15 Gy.

It can be concluded that it is possible to measure a single spot with the CCD system, a measurement which is difficult to perform with a calibrated ionization chamber (of which the effective measurement volume is in the order of magnitude of the spot size).

### 5.4.2 Dose distribution in a plane

The purpose of this measurement was, in addition to examining the CCD-screen system properties, to understand the influence of spot spacing on dose homogeneity.

The dose distribution was produced without range shifter plates and using 5\(\times\)5 spots (with a FWHM of 7.6 mm) with a distance of 5 mm in between spots in both T and U-directions. Figure 5.6A shows the dose distribution observed with the CCD system. There is a clear ripple in the dose distribution visible, which is larger in the U-direction. The magnitude of the ripple observed with the CCD system is \(\approx\) 7 %.

The asymmetry in the T and U-direction of the observed ripples may be caused by the asymmetry in the profile of the single spot (see figure 5.4). To verify this, we have performed a simulation in which single spots are added numerically to produce the dose distribution from figure 5.6A. The result is shown in figure 5.6B. The com-
Comparison shown in figure 5.6C shows that the ripple pattern can be well reproduced by the calculation.

With the aid of this ‘spot simulation’ tool we can determine the distance between spots which is sufficient to achieve a homogeneous dose distribution (ripple below 3%). The result is 4 mm. However, in that case the treatment time will increase with a factor \((1+0.2\cdot(1\text{-duty factor}))^2 = 1.17 = 17\%\). For clinically used dose distributions, there is a trade-off between spot distance and treatment time. The same holds for the spot size: by tuning the ion optics of the gantry, one is able to vary the spot size (within certain limits). By choosing a larger spot size, the homogeneity of the dose distribution will increase, but the dose fall-off distance (or penumbra), will also become larger. At PSI one has therefore chosen a compromise between spot size (7.6 mm) and inter-spot distance (5 mm) that is optimal for the dose distributions in clinical situations. It should be noted that this ripple is smeared out at larger depth, where the spot size becomes larger due to multiple scattering.
From the ripples observed in the dose distribution in figure 5.6A it can be concluded that the distance between the spots of 5 mm is too large compared to the width of 7.6 mm of a pencil beam at a depth of 1 cm polystyrene. The dose distribution can be reproduced in a simulation in which single spots are added numerically. The asymmetry in the observed dose distribution of figure 5.6 is caused by the asymmetry of a single spot dose distribution. In order to achieve a ripple below 3 % at a depth of 1 cm polystyrene, the distance between the spots (as currently used in the PSI gantry) has to be \( \leq 4 \) mm.

5.4.3 Dose distribution in a volume

The measurement of the dose distribution in a 3 dimensional volume using the 2 dimensional CCD-scintillator screen system can be performed in two ways:

- using a brute force method, in which the dose measurement is repeated for a large number of phantom depths. This method is used for the measurement of the depth-dose relation (see section 5.3). Since the effect of backscatter is small for protons (see section 3.5.4), no phantom material behind the scintillator has to be used.

- for QC purposes it is also possible to position the screen at a fixed depth, and make use of the fact that in a spot-scanning system we know the elementary pencil beam profiles that make up the dose distribution.

The purpose of this section is to study the second method. It should be realized that the spot profiles are not equal under changes of the \( S \) or \( U \) coordinate. In first approximation, however, we can use the profiles from figure 5.4 for all spot positions.

As an example we have used a \( 6 \times 6 \times 6 \) cm homogeneous, box-shaped dose distribution of 2 Gy, which is used as a reference dose distribution for the PSI dosimetry. The experimental setup is shown in figure 5.7.

The number of range shifter plates varies from 14 to 30. The spot weights (see section 5.2.1) are chosen in such a way, that the resulting dose distribution in PMMA (=lucite) is a homogeneous \( 6 \times 6 \times 6 \) cm box (analog to the Spread Out Bragg Peak, illustrated in figure 1.4 in section 1.1.3), of which the distal edge is positioned at a depth of 12.5 cm PMMA. The dose in the distal edge originates from spots which had 14 shifter plates moved in their beam path. The relative weight of these spots was 31 %, while the most proximal spot (30 range shifter plates) had a weight of 3 %. All spots with the same number of range shifter plates have the same weight, and are positioned at a distance of 5 mm. The dose at the position of the screen (at a depth of 1 cm polystyrene) is 70 % from the dose in the target volume: 1.4 Gy, according to the treatment planning calculation.

The observed light distribution at a depth of 1 cm polystyrene is shown in figure 5.8A. The dose distribution is much more homogeneous than the distribution shown in figure 5.6A, since the pencil beams are now broadened by the range shifter plates.
To investigate the error-detecting capabilities of our CCD system, we have deliberately altered the steering program of the $6 \times 6 \times 6$ cm box, so that one spot positioned at the distal side of the box (with 14 range shifter plates moved in) was not delivered. The observed light distribution is shown in figure 5.8B. At first sight the observed light distribution has not changed. However when we subtract the image with the missing spot from the complete image, see figure 5.8C, the missing spot is clearly visible.

We have also measured the dose resulting from the missing spot alone. This is shown in figure 5.8D. The mean value in a region of $7^2$ pixels in the centre of this spot was $267 \pm 9$ ADU $\equiv 2.2 \pm 0.07$ cGy (using the conversion factor for the PSI May 1997 setup\(^2\), from table 4.6: $11.7 \times 10^4$ ADU/Gy). The mean value for the same region in the difference image was $344 \pm 81$ ADU $\equiv 2.9 \pm 0.7$ cGy. These values are to be compared with the dose of 1.4 Gy at the position of the screen, which means that with this technique a missing distal spot contributing only 2% of the observed dose, can still be detected.

At the first sight the removal of a single spot in the distal plane is expected to have a variation of $0.3$ (peak plateau-ratio) * 0.34 (spot weight) = 10% at the position of the screen. However, since the spots overlap, the total dose is larger than the dose in the centre of the spot (see also section 5.4.2). For the 2 dimensional case we have shown that for an inter-spot spacing of 5 mm and FWHM of 7.6 mm the ratio between the dose in the centre and total dose is 34%. Therefore in the 2 dimensional case a decrease of 3.4% at the position of the screen is expected by removing of one spot. In the real 3 dimensional case there will also be an overlap with the spots located in the plane just in front of the distal one. This may explain the observed variation of 2% at the position of the screen.

When we also want to detect errors that are caused by the removal of proximal spots (30 range shifter plates) the situation becomes more complicated. The weight of those spots is only 3%, which means that when such a spot is removed, the variation in dose at the position of the screen is smaller than 1% (the peak-to-entrance ratio for the PSI beams is $\approx 3$, see figure 5.3). In section 4.4.1 it has been shown that the width $\sigma$ of the pixel noise distribution is 0.71% for a dose of 1.4 Gy. Therefore in its present form it is not possible to detect with sufficient certainty ($p > 0.95$) if errors smaller than 1.4% ($2\sigma$) have occurred. This means that for the detection of errors caused by proximal spots using the present system the screen has to be moved to larger depths.

The CCD system, kept at a fixed position, has the promise to be able to verify the 3 dimensional treatment plan. In order to be able to verify the spots deposited at all depths (including the most proximal), more advanced statistical techniques that make use of the correlation between the pixel values in an image are needed to deal with the noise.

\(^2\)This number is different from the one used in the previous section (November 1996) because of a different diaphragm opening and lens-screen distance.
5.4 Spot scan verification

Figure 5.7: Experimental setup for the measurement of 3 dimensional dose distributions, using the CCD-screen system at a single depth. The position and size of the desired target dose distribution are also indicated.

Figure 5.8: The detection of a missing spot in the homogeneous 6×6×6 cm box. Measurement performed at a depth of 1 cm polystyrene. A: complete box, B: without 1 spot, C: difference between A and B, D: direct measurement of the missing spot.
5.5 Measuring the effect of inhomogeneities

5.5.1 Range dilution

The high spatial resolution and large sensitivity of the CCD system also enables us to verify treatment plans which are difficult to verify with conventional dosimetry. Examples of such treatment plans are dose distributions which cross a bone-tissue inhomogeneity. For protons the effects of inhomogeneities are larger than for X-rays, because the proton energy loss is more sensitive to the material parameters than the X-ray mass absorption coefficient. For protons the change in material parameters can cause an unexpected change in the position of the Bragg peak, which can easily cause severe under- or overdosage.

Another effect which plays a role is the change in multiple scattering for different materials. Both the proton flux and proton energy loss will be affected, since protons which travelled different paths have also different energy spectra. For inhomogeneous media the effect of a distribution in proton pathlengths due to the multiple scattering process has larger consequences than for homogeneous media where the detour factor is small for protons above 1 MeV (see equation (2.20) in section 2.1.4). It also affects the dose distribution in regions which are not directly behind an inhomogeneity. The effect is called range dilution \([113, 127]\).

Most present-day treatment planning programs use pencil beam algorithms. These algorithms calculate the dose distribution by adding the dose of individual pencil beams which travel straight ahead. In such algorithms an inhomogeneity in the path of a pencil beam only affects the region behind the inhomogeneity. Pencil beam algorithms will therefore yield inaccuracies in case of range dilution. More accurate calculations are provided by Monte Carlo calculations, in which individual protons are followed. These are, however, presently too slow for routine treatment planning calculations. In a recent Ph.D. thesis \([113]\) various calculations are compared in order to estimate the effect of range dilution in clinical situations. In order to verify these calculations by measurements, a very good spatial resolution is needed. In this section we will show that this is possible with our CCD system. The purpose was to measure the effect of a water-PMMA (lucite) interface in front of a homogeneous dose distribution. The experimental setup is shown in figure 5.9.

The planned dose distribution was a homogeneous \(4\times4\times4\) cm box at the position behind the \(\Gamma\)-shaped water/PMMA interface (see figure 5.9). The beam weights of the spots with different range shifter plate settings have not been corrected for range dilution, i.e. are calculated with the conventional spot scanning treatment planning system in such a way that a homogeneous dose distribution is obtained. We measured the 3 dimensional dose volume using the brute force method of section 5.4.3: that is by adding a variable amount of PMMA slabs in front of the scintillator screen. Since the effect of backscatter is small for protons (see section 3.5.4), no phantom material behind the scintillator was used. In figure 5.9 the measured dose distributions both at the begin-
5.5 Measuring the effect of inhomogeneities

ning of the box (1 cm PMMA slabs) and at the end of the box (4 cm PMMA slabs) are shown.

There is no clear effect of the inhomogeneity at the beginning of the box. At a depth of 4 cm however, which is at the end of the 4×4×4 cm box, an effect is visible. There is decrease of dose at the region of the interface at the side of the PMMA. Since PMMA is more dense (ρ=1.16 g/cm³), more protons will scatter out of the PMMA into the water than vice versa. Therefore the proton flux, and thus the dose, at the end of the range decreases.

Pencil beam algorithms in which only local effects are taken into account would yield wrong results, especially in the neighborhood of the Bragg peak, where the contribution to the dose is largest. In [113] it shown that the measured results from figure 5.10 are qualitatively in agreement with MC calculations.

5.5.2 Segmented beam monitor

Another effect which is caused by the multiple scattering processes has been observed in a measurement with 80 MeV proton beams from the facility in Louvain-la-Neuve. In order to verify the beam centering and flatness, the beamline in Louvain-la-Neuve contains a segmented beam monitor, which is positioned at a distance of 30 cm from the isocentre. This monitor is an ionization chamber, from which the foils consist of 4 segments of thin layers (≈ 50 μm) gold evaporated on mylar. The gold segments are separated by a small (≈ 1 mm), empty zone which contains only mylar. The differences in energy loss for protons passing through the plastic+gold and protons passing through just the plastic are very small, due to the small thickness of the gold layer. Still the structure of the segmented beam monitor was clearly visible, even at small depths (1 cm polystyrene), as is shown in figure 5.11.

This can be explained by the multiple scattering: protons that pass through the gold have larger deflection angles than protons that pass only the plastic. Therefore at the position of the plastic gap, there are in addition to the protons that passed directly through plastic also protons which are scattered from the gold. At the position of the gold segments, however, there is just the contribution from protons that passed the gold, since much less protons are scattered by the plastic towards the gold segments. Therefore at the position of the plastic gaps the proton flux is increased, and therefore also the observed dose in the screen. The observed maximal effect is 2.7 %. In the one dimensional profile, it is difficult to see. In a two dimensional image, however, it is much clearer because of the spatial correlation of the effect. This again demonstrates the usefulness of a high spatial resolution and large sensitivity 2 dimensional measurement system.

Qualitatively the effect can be understood using equations (2.18), (2.19) from section 2.1.4. For 80 MeV protons the mean scattering angle θ₀ for a 0.1 mm thick plastic foil (effective thickness 0.1 g/cm²) is ≈ 4 mrad. For a gold layer of 50 μm (effective thickness also 0.1 g/cm²) the θ₀ becomes 9 mrad. At the position of the screen (30 cm)
Figure 5.9: Top view of the experimental setup of range dilution experiment. The planned dose distribution was a homogeneous $4 \times 4 \times 4$ cm box behind the $\Gamma$-shaped water/PMMA inhomogeneity.

Figure 5.10: Results of the range dilution experiment (see figure 5.9). The dose distribution is measured with the CCD system at a depth of 1 cm (A) and at 4 cm depth (B). The PMMA is positioned at the right side. C and D are profiles at the position of the lines. The dotted line through the profiles indicates the position of the transition water-PMMA.
this means that 30% of the protons that pass the gold are scattered over a distance of more than 2.5 mm. The proton flux will therefore increase at the position of the gap, since there are contributions from both the protons that went through the gap, and protons that are scattered by the gold.

5.6 Conclusions and outlook

In this chapter we have shown that the CCD-scintillator system, is a useful tool for the quality control of scanning proton beams, such as delivered by the PSI spot-scanning gantry.

With respect to the dosimetric properties of the system, the ratio of measured/calculated yield is in agreement with the values obtained in a passively scattered beam, which means that no dose-rate effect can be observed. The reproducibility between two measurements performed at PSI which were separated in time by 6 months was within 1.5%.

With respect to the decrease of the light signal in the Bragg peak compared to the signal in the entrance region (at 1 cm depth), the effective $k_B$ value for the PSI
beam was determined to be 4.0 mg cm\(^{-2}\) MeV\(^{-1}\). This yields a decrease of 5 % in the Bragg peak, which is smaller than in the Uppsala beam: 8 %. This can be qualitatively explained by the larger initial energy spread of the PSI beam.

The CCD-scintillator system allows detailed studies of the homogeneity of the dose distribution in a plane. It was found that at small depths (1 cm polystyrene) an inter-spot distance of 5 mm is too large for the spots currently used in the PSI gantry (FWHM = 7.6 mm), since it yields a ripple of 7 %. We have developed a computational tool which simulates the addition of spots. Using this tool it was found that a spot distance of 4 mm is necessary to achieve a homogeneous dose distribution at these small depths. However a 7 % ripple at such a small depth is no problem since the total dose is much smaller anyhow at low depths.

A quick verification of a 3 dimensional dose distribution is possible using CCD measurements with the screen at a fixed depth. This was shown in an experiment in which deliberately a spot was removed, from a steering program that was designed to produce a homogeneous 6×6×6 cm dose distribution. The Bragg peak of the removed spot was positioned at the distal side of the box. Its contribution to the total dose at the position of the screen, located at 1 cm depth from the entrance was 0.03 Gy (2 % of the total dose). This turned out be easily detectable.

In the second part of this chapter we used the CCD-scintillator system to verify a treatment plan that contains an inhomogeneity. The small dose variations caused by differences in multiple scattering in different materials can be measured accurately.

**Outlook**

The use of the CCD-scintillator system will be increased, if in the treatment planning software a correction for the light decrease in the Bragg peak is included. In this way instead of *isodose* contours, the software can calculate *isolight* contours. This will facilitate direct comparison of the treatment plan with the measured result. The light decrease correction has to be determined for each specific beam quality.

The results described in this chapter indicate a possible extension of the use of CCD+screen system: namely as an *in vivo dosimeter*. The screen can mounted on the skin of the patient (or just above) with the scintillating side on top. The minor effect this will have on the proton dose in the patient, can be corrected for by the treatment planning system. The CCD camera can be positioned next to the gantry head, so that it observes the screen under an angle. Because of this angle the image will be geometrically distorted, but this can easily be taken into account. The CCD camera can now record on-line the applied dose to the patient. In this way a device is obtained that has the same function as a portal imaging system (see section 1.2.3): an independent quality control on the delivered irradiation. Since conventional portal imaging as used for X-rays is not possible for protons, due to the stopping of protons in the patient, this may turn out to be a very promising method.