

# A REAL-TIME BEAM MONITORING SYSTEM FOR HIGHLY DYNAMIC IRRADIATIONS IN SCANNED PROTON THERAPY: DERIVATION OF SAFETY TOLERANCES\*

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## Abstract

Patient treatments in scanned proton therapy exhibit dead times, e.g. when adjusting beamline settings for a different energy or lateral position. On the one hand, such dead times prolong the overall treatment time, but on the other hand they grant possibilities to (retrospectively) validate that the correct amount of protons has been delivered to the correct position. Efforts in faster beam delivery aim to minimize such dead times, which calls for different means of monitoring irradiation parameters. To address this issue, we report on a real-time beam monitoring system that supervises the proton beam position and current during beam-on, hence while the patient is under irradiation. For this purpose, we sample 1-axis Hall probes placed in beam-scanning magnets and plane-parallel ionization chambers every 10  $\mu$ s. FPGAs compare sampled signals against verification tables – time vs. position/current charts containing upper and lower tolerances for each signal – and issue interlocks whenever samples fall outside. Furthermore, we show that by implementing real-time beam monitoring in our facility, we are able to respect patient safety margins given by international norms and guidelines.

## PREFACE

A detailed report on software and firmware enhancements of the control system has been submitted as an ordinary paper [1]. In these proceedings, we concentrate on one part of the work and derive applicable safety tolerances supervised by the real-time monitoring system.

## INTRODUCTION

In scanned proton therapy, we use a Gaussian-shaped beam of protons to irradiate cancerous tissue. The beam size  $\sigma$  in air amounts to a few millimeters. To cover the entire extent of the three-dimensional tumor volume with protons, the beam needs to be scanned transversally and in depth. At the Paul Scherrer Institute (PSI), we installed a dedicated super-conducting cyclotron that provides a continuous and mono-energetic proton beam of 250 MeV [2]. We realized transverse scanning with a pair of beam-deflecting dipole magnets; to scan the proton beam in depth, we change its energy (and, thus, penetration depth) by inserting variable amount of degrading material into the beamline [3].

The beam scanning process requires a discretization of the tumor volume: it is cut in slices of equal energy (or

penetration depth) and a rectilinear scan grid of fixed transverse (or lateral) beam positions is imposed on all of those slices. In our second-generation treatment room at PSI, we require  $\sim 100$  ms to change the beam energy between slices and  $\sim 3$  ms to scan the beam from one transverse grid point to the next [4]. The beam is turned off completely during those transitions. We use this dead time, especially the latter, to validate that the correct amount of protons has been applied to the correct position. If the deviation between measurement and expectation exceeds a certain tolerance, we have the possibility to interrupt the treatment of the patient to investigate the source of uncertainty. International norms [5] and guidelines [6] demand such frequent checks to guarantee patient safety.

At PSI, we treat patients successfully using this discretized beam scanning technique since 1996. To maximize irradiation performance and possibly broaden the window of treatable indications, we pursue implementing a faster form of beam delivery, which we call line scanning [7]. In line scanning, the beam is moved continuously along straight lines in the transverse plane giving up the idea of the fixed grid in this dimension. The 3 ms dead times are reduced to changes between lines, which yields increased performance but, at the same time, fewer opportunities for validation checks.

To provide adequate safety measures for line scanning, we introduced a dedicated beam monitoring system. We reported on its design [8] and implementation [1] in previous works. A major enhancement with respect to the conventional monitoring approach is its real-time character: we compare the measured beam position and proton deposition to predefined tolerances every 10  $\mu$ s. This cyclic comparison runs during beam-on, hence while lines are scanned. As such, we can react to errors in different beam delivery units very fast and issue beam-off commands rapidly in case of unforeseen inaccuracies or failures.

The scope of this paper is to provide a full derivation of our line scanning safety tolerances. We will focus on acceptable over/under-exposure of the healthy/malignant tissue to radiation and acceptable deviations in the transverse beam position. Based on our experience in patient treatments, we presuppose that errors in beam delivery occur rarely (less than once throughout the entire course of the treatment) and randomly. They cannot be linked to specific configurations of the machine and do not lead to any systematic effects.

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## DERIVATION OF SAFETY TOLERANCES

### Definition of Erroneous Delivery

An average treatment schedule at PSI foresees 30 irradiation sessions. In each session, an average dose of 1.818 Gy is delivered homogeneously to the tumor volume. We consider an irradiation erroneous if the homogeneity (or uniformity) across the target is significantly impaired. We set the threshold at absolute deviations of  $\pm 2\%$  or  $\pm 36$  mGy. Localized over- or underdosages of this magnitude are physically measurable, but we do not expect them to have any consequence on the clinical outcome of the treatment. Hence, we aim to restrict all excess or missing doses to  $\pm 36$  mGy, even in case of severe errors of the delivery system. Line scanning under such monitoring conditions can be considered safe.

With this definition we follow the guidelines of the International Commission on Radiation Units and Measurements. They claim that safe and accurate irradiations apply to doses that “are delivered throughout the target volume with sufficient uniformity (better than  $\pm 2.5\%$ )” [6]. Furthermore, we are well within the tolerance given by the International Electrotechnical Commission:

“The secondary dose monitoring system shall be set to terminate irradiation before an additional absorbed dose of 10% or 0.25 Gy, whichever is greater, is delivered.” [5]

In order to fulfill the  $\pm 36$  mGy constraint – even in case of severe failures of the beam delivery system – we supervise the following three quantities in real-time during patient irradiation: (i) total dose deposition, (ii) instantaneous beam current, as well as (iii) transverse beam position. We measure the former two quantities with two independent parallel-plate ionization chambers [9] that are connected to cyclic readout electronics (data samples every 10  $\mu$ s). They are the final diagnostic elements in the beamline and placed between the exit window of the vacuum beam pipe and the patient. The beam position in the transverse plane is supervised indirectly by one-axis Hall probes, which measure the field strength of the beam-scanning magnets at identical readout frequency. High field strengths correspond to a large transverse deflection of the proton beam and vice versa.

In the following sections, we will derive acceptable safety tolerances for all three quantities. We base our derivation on the assumption that all machine-related errors occur rarely. Hence, we regard a combined failure of multiple elements as highly improbable.

### Maximum Dose Deposition of a Proton Beam

When protons penetrate through matter, they continuously lose energy before coming to rest. The energy loss is low at the beginning of their path and reaches a maximum near the end of their range. Due to collisions with target nuclei, protons may experience large angle deflections. The combined effect of characteristic energy deposition and transverse scattering can be seen in Fig. 1.

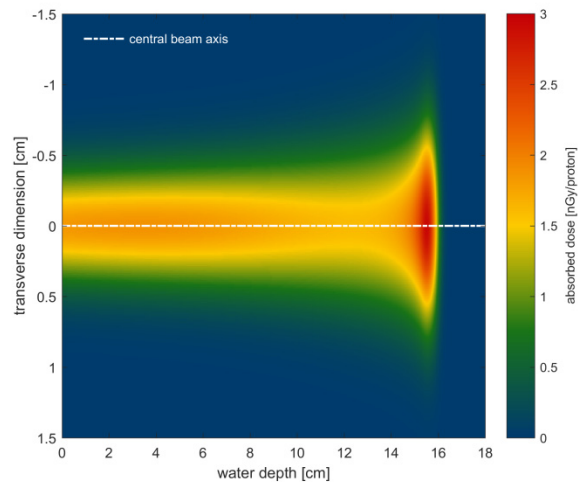


Figure 1: Energy deposition of a 150 MeV proton beam in water. The initial beam width  $\sigma$  was assumed to be 2.9 mm. The calculation was performed using the analytic transport models of the PSI treatment planning software.

The energy deposition is quantified as absorbed dose  $D$  and formally defined as

$$D = \Delta \epsilon / \Delta m, \quad (1)$$

with the mean energy imparted  $\Delta \epsilon$  in the mass element  $\Delta m$ . Doses can be measured in finite volumes  $\Delta V$  only. The PSI treatment planning software uses a voxel size of

$$\Delta V = (4.0 \times 4.0 \times 2.5) \text{ mm}^3. \quad (2)$$

If we place such a small volume on the central beam axis (white dashed line in Fig. 1) and calculate the average dose deposition as a function of penetration depth, we obtain the ‘on-axis’ Bragg curves shown in Fig. 2.

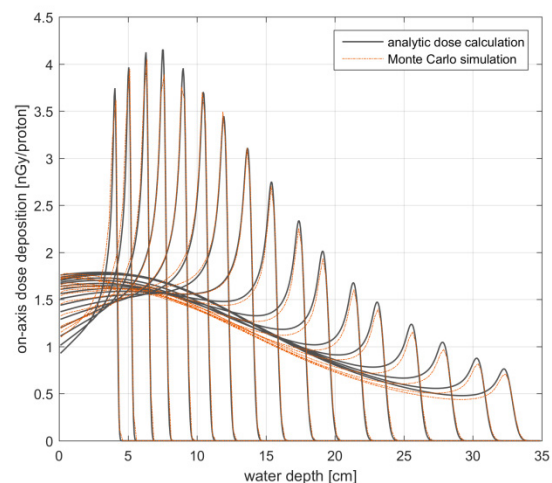


Figure 2: Analytically calculated vs. Monte-Carlo-simulated on-axis Bragg curves for 17 different incident energies (70, 80, ..., 220, 230 MeV).

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We see that the maximum energy deposition never exceeds

$$D_{\max} = 4.2 \text{ nGy/proton.} \quad (3)$$

Furthermore, the analytic calculations [10] agree with the Monte Carlo simulations [11], which are based on a detailed phase space model of our proton beam. We conclude that, independent of the beam energy, we will never deposit more dose than  $D_{\max}$  inside a small volume of size  $\Delta V$  assuming that the material contained is water-like.

### Tolerance for Dose Deposition

Both ionization chambers have been calibrated to measure the proton beam current  $I$ . Internally, they operate on monitor units (MU), which can be correlated to absolute number of protons using the calibration curve shown in Fig. 3. The maximum authorized beam current in the treatment room is limited to 500 MU/ms. For example, at 120 MeV we find

$$I_{\max} = 500 \text{ MU/ms} \approx 500 \text{ pA.} \quad (4)$$

Scaling all on-axis Bragg curves of Fig. 2 by their corresponding MU-conversion factor (see Fig. 3) yields a maximum dose deposition of

$$D_{\max} = 0.024 \text{ mGy/MU.} \quad (5)$$

Hence, we can set the tolerance for the deposited dose to

$$\delta D_{\max} = \frac{36 \text{ mGy}}{D_{\max}} = 1'500 \text{ MU,} \quad (6)$$

which means that the delivered dose distribution will have sufficient homogeneity if deviations are within  $\pm \delta D_{\max}$ .

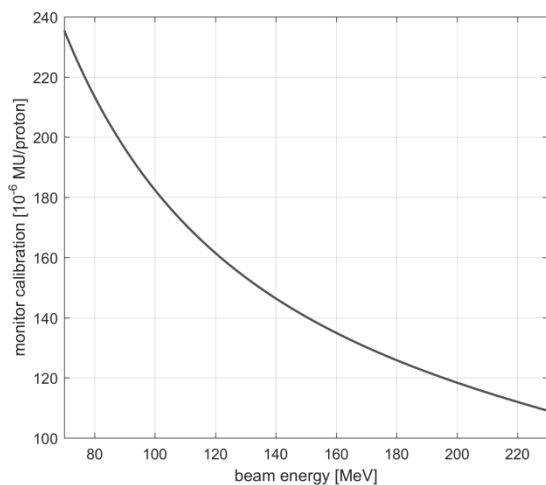


Figure 3: Calibration curve for the primary ionization chamber correlating internal monitor units (MUs) to absolute number of protons for different beam energies.

### Tolerances for Instantaneous Beam Current

From Eq. 4 and Eq. 6 we can immediately calculate the tolerance for fluctuations in the beam-on time:

$$\delta t_{\max} = \frac{\delta D_{\max}}{I_{\max}} = 3 \text{ ms.} \quad (7)$$

In addition, we would like to specify a margin for the instantaneous beam current to define a tolerance band similar to the one depicted in Fig. 4 below. In a worst case error scenario, the beam current would follow the red arrow in Fig. 4: It would increase by  $\delta I$  at the very end of the irradiation and remain on for  $\delta t$  although we would expect zero beam current at that point. The excess dose must be below 1'500 MU, hence

$$(I_{\max} + \delta I) \delta t \leq 1'500 \text{ MU.} \quad (8)$$

We can fulfill this criterion, by choosing a  $\delta I$  of 50 MU/ms and a  $\delta t$  of 2 ms. Note that we chose a smaller value here for the tolerance  $\delta t$  compared to Eq. 7, since we need to consider the time it takes to switch off the beam after a violation of an error band has been detected (typically  $\sim 0.3$  ms).

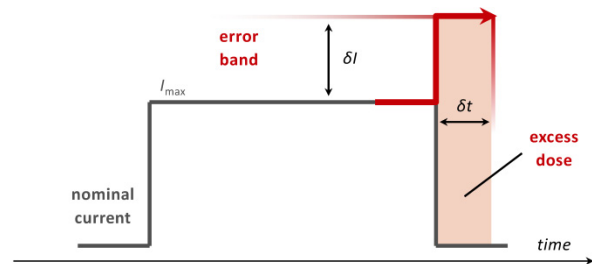


Figure 4: Schematic of the upper beam current tolerance band comprising margins in time ( $\delta t$ ) and current ( $\delta I$ ).

### Tolerance for Beam Position

The homogeneity of dose distributions is very sensitive to misplacements of the beam in the transverse plane. Small inaccuracies can yield areas of significantly higher and lower dose concentration with respect to the prescription. As such, the supervision of the transverse beam position requires a strict tolerance.

Let us assume a square ( $10 \times 10 \times 10$ )  $\text{cm}^3$  target volume filled with water that we plan to irradiate to 1.818 Gy. A simple treatment plan could foresee 25 different iso-energy slices, each filled with straight lines delivering uniform dose. Fig. 5 below visualizes this example case. If we assume a 1 mm deviation in position that is localized within a single dose calculation voxel  $\Delta V$ , the resulting differences in planned and delivered dose distribution will be of the order of  $\pm 30$  mGy (without derivation). Hence, we conclude that errors in the transverse beam position should be smaller than  $\pm 1$  mm.

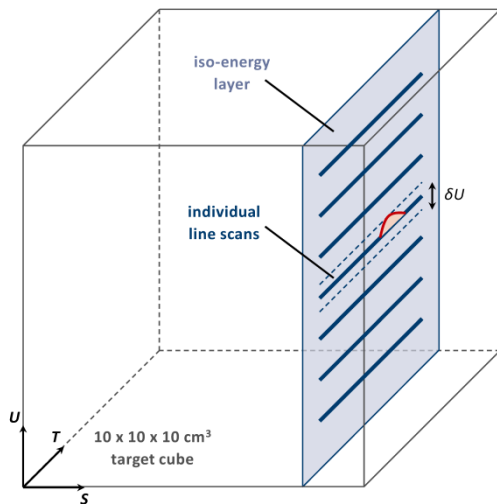


Figure 5: Example for an error in the transverse beam position (deviation from the planned beam position indicated in red).

## DISCUSSION OF SAFETY TOLERANCES

We have derived a set of beam monitoring tolerances for line scanning irradiations in proton therapy from fundamental dosimetric considerations. These tolerances, combined with effective safety measures that terminate the irradiation in case of errors, facilitate complying with safety constraints given by international norms and guidelines. Although actual tolerance values may be specific to line scanning irradiations using a proton beam, we are confident that they could be derived in a very similar way for different particles (e.g. helium or carbon ions) and different beam scanning techniques (e.g. spot or raster scanning).

### Definition of Erroneous Delivery

In the definition of an erroneous delivery we use an absolute value of  $\pm 36$  mGy. In our patient history we have very few cases, in which the prescribed dose for a single session is below 1.818 Gy. Thus, we can satisfy the  $\pm 2\%$  criterion for almost all patients. Numerous studies actually suggest that a treatment with fewer sessions and higher dose per session could be advantageous for radio-resistant tumors, especially when irradiating with carbon ions. In such cases, the  $\pm 36$  mGy threshold would actually correspond to a relative tolerance smaller than  $\pm 2\%$ .

### Maximum Dose Deposition of a Proton Beam

The height of the on-axis Bragg curves in Fig. 2 depends on the voxel size  $\Delta V$  and the contained material. The average dose deposition increases with increasing material density and decreasing voxel size. For infinitesimally small volumes  $dV$ , the value for the maximum dose deposition amounts to  $\sim 5$  nGy/proton. Nevertheless, we would argue that a finite volume of  $40 \text{ mm}^3$ , as used in our derivation, is well below typical sensitive volumes of standard ionization chambers (e.g.  $600 \text{ mm}^3$  for PTW Farmer chamber

type 30010). And we think that all tolerances should be defined based on quantities that are actually measurable with standard dosimetry equipment.

### Tolerance for Dose Deposition

We regard the tolerance  $\delta D_{\max}$  (see Eq. 6) as an important tool against systematic over or under-dosage (e.g. beam current in Fig. 4 constantly too high or too low). To verify it during the irradiation of a line, we added an integration unit in the readout electronics of one of our parallel-plate ionization chambers. As such, it monitors the total number of delivered MUs (or protons) as function of time rather than the instantaneous beam current. Hence, we can immediately abort the irradiation whenever a violation of  $\delta D_{\max}$  has been detected. Dedicated firmware features of this tool are described elsewhere [1].

### Tolerance for Instantaneous Beam Current

We use the remaining ionization chamber to monitor the instantaneous beam current  $I$ . The noise level of the chamber mainly determines the value for  $\delta I$ . On the one hand, it is desirable to decrease  $\delta I$  as much as possible to guarantee high safety standards, but, on the other hand, a low  $\delta I$  can become very impractical if the treatment of a patient is interrupted too frequently.

From Eq. 8 we see that  $\delta t$  is directly correlated to the maximum authorized beam current  $I_{\max}$ , with which patients can be irradiated. Efforts in decreasing the irradiation times even further often suggest increasing  $I_{\max}$  (by roughly a factor 10). To guarantee the same level of safety,  $\delta t$  would need to be lowered in that case. We think that a value of  $200 \mu\text{s}$  could still be practical considering all delays and reaction times of our beamline and controls.

### Tolerance for Beam Position

In our opinion, the tolerance for the transverse beam position ( $\pm 1$  mm) represents the strongest constraint. Because therapeutic beam currents are generally low compared to many experimental beamlines, signals in position-sensitive monitors often suffer from high noise levels. Furthermore, these monitors need to be as non-destructive as possible in order to avoid deteriorations in the beam quality. As such, their spatial resolution is limited to  $\mu\text{m}$ -thin readout channels every one or two millimeters. Following the ALARA principle (as low as reasonably achievable), many centers increase their position tolerance (e.g. to  $\pm 1.8$  mm at MD Anderson Cancer Center, Houston, TX, USA [12]). Other viable approaches are adapting the tolerance according to the beam energy (and, thus, beam size in air) or to use indirect measures on beam position such as the field strength of the scanner magnets (our approach for line scanning). The latter comes at the advantage of exhibiting a stable signal-to-noise ratio independent of the proton beam current, but it provides an indirect measurement of the transverse beam position only.

## CONCLUSION

We provided a set of tolerances and their derivation for line scanning irradiations in proton therapy. We consider patient treatments in line scanning mode safe, when supervising those tolerances frequently (in our case every 10  $\mu$ s) during the irradiation.

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