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ON-LINE DYNAMIC BEAM INTENSITY CONTROL IN A PROTON THERAPY CYCLOTRON*

Serena Psoroulas, Pablo Fernandez Carmona, David Meer, Damien Charles Weber[1]
Paul Scherrer Institut, 5232 Villigen PSI, Switzerland
[1]also at University of Zurich, Zurich, Switzerland, and University of Bern, Bern, Switzerland

Abstract

Modern proton therapy facilities use the pencil beam scanning (PBS) technique for the treatment of tumours: the beam is scanned through the tumour volume sequentially, i.e. stopping the beam at each position in the tumour for the amount of time necessary to deliver the prescribed dose for that position, and then moving to the next position (dose-driven delivery). This technique guarantees robustness against fluctuations in the beam current. Modern cyclotrons however offer very stable beam currents, and allow regulating the beam intensity online to match the requested intensity vs. time profile of the beam ('time-driven' delivery). To realise time-driven delivery at the COMET cyclotron at the Paul Scherrer Institute (PSI), we have designed a beam intensity controller which is able to partially compensate for the non-linearity and the delay introduced by the physical limitations of the beam line elements and its drivers; this is particularly important when trying to achieve a very fast modulation of the beam, as required by clinical plans. Experimental results have shown good performance for most current clinical scenarios, and we are investigating more advanced solutions for higher dose rates scenarios.

INTRODUCTION

Proton therapy (PT) is a radiation therapy technique which established itself recently as treatment of choice of many tumours, particularly paediatrics [1]. Modern PT facilities use the pencil beam scanning technique (PBS) for the treatment of deep-seated tumours, because it provides better tissue sparing and less neutron contamination than other delivery techniques, such as passive scattering. In PBS, the beam is moved sequentially through the target volume, and stopped at each point through the volume for the amount of time needed to deliver the amount of protons defined by the treatment plan (therefore is also called ‘discrete scanning’). About 75% of all PT facilities feature a cyclotron [2], as this technology offers high intensities and a very reliable beam current output, which are both advantageous to keep treatment times within predefined limits (about 2 Gy/minute needed to irradiate a 1-liter volume). Despite this excellent timing performance, PBS is currently mostly used to treat static tumours (for example, brain tumours) and in some cases used to treat tumours with limited periodical motion due to respiration, such as lung or liver, with or without motion mitigation strategies. This is due to the fact that the reciprocal/independent motion of the beam and the target cause an interference pattern in the resulting dose distribution, that worsens the delivered dose distribution (the so-called the interplay effect) and makes in the end the treatment ineffective. PBS can be used to treat such tumours only in combination with motion mitigation techniques [3], to ensure the dose degradation remains within acceptable limits.

At PSI, in the clinical treatment unit Gantry 2, we are investigating a new delivery technique, continuous line scanning (CLS), which offers substantially lower treatment times and better dose conformity of moving targets treatments, particularly when combined with motion mitigation. In recent papers [4, 5] we compared it with the two main discrete scanning techniques used for PBS, the so-called spot scanning [6] and raster scanning [7], showing that CLS brings clear advantage over the other techniques for liver tumours treated with motion mitigation.

One of the keys of CLS performance is its high flexibility and speed in the dose modulation, which is achieved by both quickly adapting the beam transverse scanning speed as well as beam intensity to what is specified for the treatment. This however requires fast intensity changes to be performed already at the cyclotron. Though the PSI COMET cyclotron is designed to match such a requirement, achieving a reliable intensity control that also meets the stringent safety requirements for patient routine treatments represent a challenge not fully considered at the time of design. We have preliminarily reported [8] a first attempt at designing a beam intensity controller for this application, and the challenges of the final design and implementation [9]. After summarising the main challenges and the characteristics of our design, we will report in this paper the experimental validation of the implemented solution.

BEAM INTENSITY CONTROL AT THE PSI PROTON THERAPY CYCLOTRON

The COMET Cyclotron at PSI and its Intensity Regulation

The COMET cyclotron (ACCEL/Varian) [10] provides a beam of 250 MeV to the treatment rooms at PSI. The beam energy is then lowered to what needed for the treatments in a degrader and energy selection section placed downstream. The beam intensity is defined at the cyclotron and gets considerably lowered when passing through the energy selection system. For keeping the beam delivery efficiency high, stable beam currents and a high cyclotron output are of utmost importance.

Inside the cyclotron, the beam is extracted from a cold-cathod-type proton source by a puller, and then accelerated

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passing through four dees. The source output is tuned once a day and kept stable during operation [11], while the regulation of the beam current is done in the next turn, by passing the beam through a vertical electrostatic deflector (VD) followed by collimators [12]. When the VD is charged, it deflects the beam off the central plane. This will cause a part of the beam to be collimated in the collimators placed in the following dee. The larger the VD electric field, the more protons will be stopped. With such a system, we can achieve intensity variations from 0 to maximum current within 50 μs.

Beam Intensity Regulation Challenges

VD voltage versus current relationship The beam current is constant at the treatment room. However, because of the losses caused by the energy selection, which are higher for lower energies, the beam output requested at the cyclotron depends strongly on the energy to be delivered at the patient. This means that we need a large range of VD voltages during treatments, with the highest values required for higher energy treatments, and values close to 0 required for low-energy treatments.

As shown in Fig. 1, to achieve a current of 1e6 protons/ms at the patient, we may require about 0.4 kV at 150 MeV, but only 0.05kV at 70 MeV. Additionally, because of fluctuations in the current extraction efficiency, ion source operation etc., the relationship between VD voltage and current at the patient can strongly vary between different days and even within the same day [8]. To better correct for such fluctuations, we perform measurements of this relationship regularly through the day, and use this information to make a first estimate of the VD voltage operating point requested by each line before delivery.

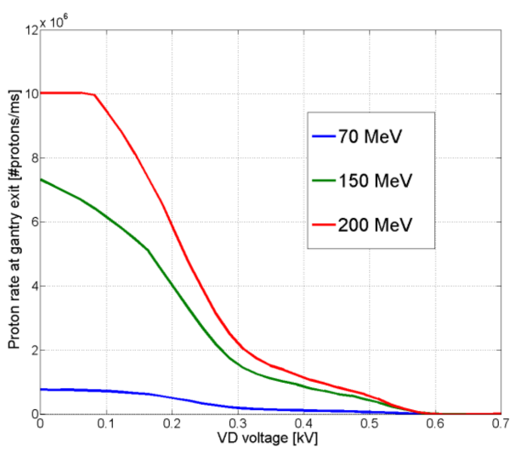

Figure 1: Vertical deflector (VD) voltage vs beam intensity at patient relationship for three energies [3].

Reaction time of the monitoring system The beam current delivered at the patient, inside the treatment room, defines the reference current for the regulation. This however introduces a systematic delay, caused respectively by the time needed for the communication between our treatment control system and the VD power supply, the time needed for the beam acceleration inside the cyclotron and the transport to the treatment room, and finally the time needed for the measurement of the beam intensity in the Gantry 2 monitoring system (which uses ionisation chambers with a collection time of about 90 μs). In total, the latency caused by the delays amounts to about 200 μs. In clinical practice, the smallest lines we want to deliver are about 300 μs long. This makes the regulation of the beam current for such lines particularly challenging.

Power supply hardware constraints Due to the harsh radiation environment in the cyclotron bunker, the power supply has been installed outside. The long cable thus required causes adds a capacitance of approximately 5 nF to the load of the system; in comparison, the capacitance due to the copper plates of the VD is only of the order of 100 pF. Together with the producer of the power supplies [13], we have optimised the impedance matching between the power supply and the load to avoid problems as much as possible. However, such a high capacitance presents a limitation to the highest speed of intensity change reachable by the power supply. Furthermore, the power supply is internally built using switched stages of 90 V, which makes the internal regulation faster but causes unpredictable overshoots to appear whenever the power supply switches stages [9].

Intensity Controller Design

We considered the different challenges identified in the previous section in the design of the intensity controller [9]. From data collected over a year of operation we realised that the large variability of the system as well as the presence of large overshoots (particularly for low voltage variations) required different parameters for the controller, depending on the expected range of possible overshoots. Therefore, our implementation foresees the presence of three different controllers, with different degrees of reaction time and robustness (the slower the controller to reach the set point, the more robust and less prone to oscillations). Each controller is better suited for different levels of overshoot and voltage variations. All three controllers have the same structure, only different parameters. During beam irradiation, the Gantry 2 control system selects the desired controller parameters based on the intensity variation (the 'gain' caused by the voltage variation) requested. Furthermore, a Smith predictor has been integrated in the controller design to cope with the 200 μs latency. We refer the reader to another paper [9] for a detailed report of the design and implementation of the intensity controller in our beam delivery FPGA.

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RESULTS

Figure 2 shows the comparison between the old version of the controller (whose behavior was explained in a previous report [8]) and the new version. It also highlights the difference between a controller without Smith predictor and a controller with Smith predictor. The latency compensation introduced by the predictor reduces the settling time, without causing instabilities in the system. Based on these results, we concluded that the delay compensation brings substantial improvements and therefore is a fundamental part of our design, despite adding complexity to the system. However, the predictor is only as good as the quality of the model of the plant. For this reason, we will commission the controller over a long period, as we have particularly observed seasonal variations in the behaviour of the facility.

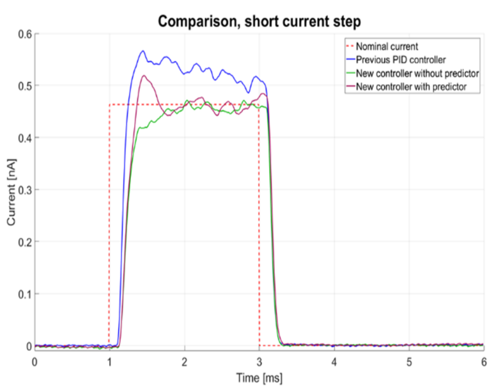

Figure 2: Comparison between controller with and without Smith predictor [8].

Concerning robustness, we tested two different scenarios: in the first case, we derived controller parameters which allowed a faster reaction time (but were less robust with respect to instabilities), while in the second scenario we derived controller parameters achieving a slower but more robust operation. We indeed could observe the faster controller become unstable after a few hours of operation, particularly when variations on the current output of the cyclotron (such as extracted beam current and transmission to the gantry room) occurred. An example of such behavior is shown in Fig. 3. From our clinical experience, we know such variations can occur almost unexpectedly during clinical operation (depending on ion source performance worsening, beam centering in the central region of the cyclotron, and other effects currently under investigation). At PSI, we are looking into possible ways to automatically tune the accelerator settings to stabilize these variations. However, since we are currently running without an automatic correction of such effects, for the time being we need to rely on a more robust controller, and therefore need to accept some compromises on the maximum settling time for the beam current in line scanning plans. Another possibility being investigated is adapting the controller parameters to the beam conditions; this solution though is not currently preferred, since it adds complexity to the Gantry 2 operation.

We could also observe the effect of the 90 V stage switching on the stability of the delivery; an example is reported in Fig. 4, where several small beam current overshoots/undershoots can be seen. In this example, one such overshoot is high enough to go beyond our safety tolerance at the beginning of the delivery, where it would have caused a beam intensity interlock. To solve this problem, we are considering the option of a linear regulation of the power supply (without discrete steps) for a future upgrade of the facility.

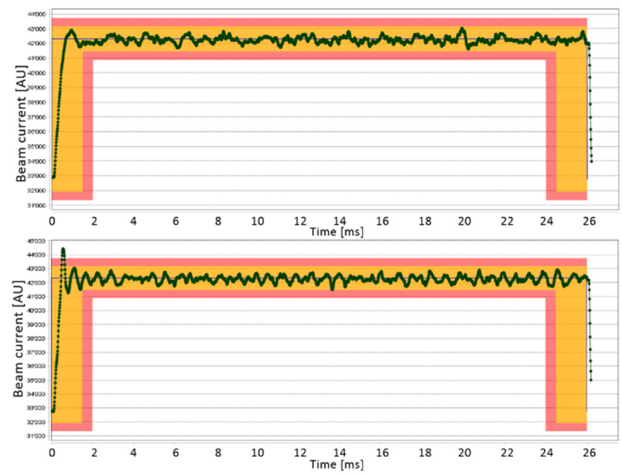

Figure 3: Beam current plots; the target current (horizontal blue line) is overlaid to the measurements (markers). The red bands represent the warning level, exceeding such bands would trigger an interlock when operating the system in clinical mode. In the example, we show the same patient file, delivered once and after one hour, with the same controller parameters; in the second irradiation, instabilities in the beam delivery arise, due to variations in the current output of the cyclotron.

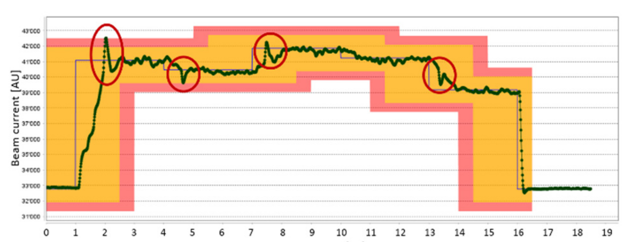

Figure 4: Effect of the 90 V stage switching: small overshoots appear during the delivery of the line.

Despite the compromises mentioned above, we could verify that the precision of the delivery meets the requirements of line scanning experiments. One example of a dose distribution delivered with the current controller is shown in Fig. 5. In this example considered, the beam current is quickly lowered to 0 and then again to maximum towards the middle of the line. The good agreement of the delivered dose distribution and the expected dose distribution is an indirect confirmation that the fast beam current modulation at the vertical deflector is working as expected. However, further testing (particularly regarding

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robustness and reliability) are necessary before the technology will be fully integrated in clinical practice.

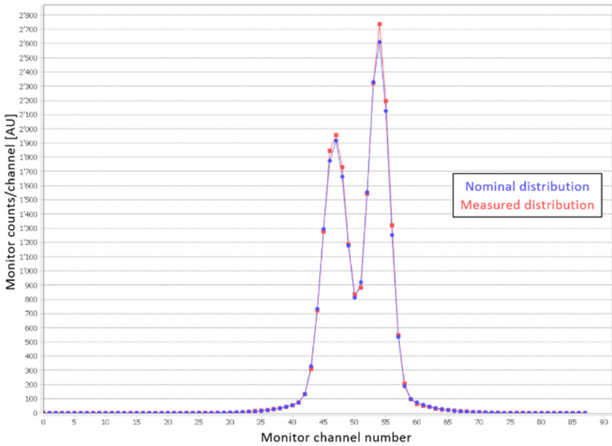

Figure 5: Dose distribution for a single line: comparison between expected ('nominal') and measured dose distributions, measured in the nozzle monitor strip chamber.

CONCLUSIONS

We reported the results of the implementation of an on-line dynamic beam intensity controller for CLS at a proton therapy cyclotron. Our design features a gain-scheduled controller, which, depending on the intensity variation required for the treatment, selects the controller parameters according to three possibilities: from slowest but more robust, to fastest but less robust to fluctuations, depending on the expected range of possible overshoots. Furthermore, to compensate for the large latency in our system, it features a Smith predictor. We have shown that the presence of the predictor substantially increases the speed of the controller. However, the large variability of the accelerator output as well as some features of the VD power supply design pose still substantial challenges to the reaction times of the system, and therefore for a first implementation we chose a more robust, though slower, version of the controller. In this implementation, currently available for experiments, most of the patient plans from our center can be delivered with a precision of few percent; however, due to the speed of the controller chosen, the settling time of the beam current might still come short of the requirements for patient cases we would like to investigate in the future. For this reason, we are planning a further upgrade of the hardware (including new power supply specifications).

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