Energy spectrum control for modulated proton beams

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In proton therapy delivered with range modulated beams, the energy spectrum of protons entering the delivery nozzle can affect the dose uniformity within the target region and the dose gradient around its periphery. For a cyclotron with a fixed extraction energy, a rangesifter is used to change the energy but this produces increasing energy spreads for decreasing energies. This study investigated the magnitude of the effects of different energy spreads on dose uniformity and distal edge dose gradient and determined the limits for controlling the incident spectrum. A multilayer Faraday cup (MLFC) was calibrated against depth dose curves measured in water for nonmodulated beams with various incident spectra. Depth dose curves were measured in a water phantom and in a multilayer ionization chamber detector for modulated beams using different incident energy spreads. Some nozzle entrance energy spectra can produce unacceptable dose nonuniformities of up to $\pm 21\%$ over the modulated region. For modulated beams and small beam ranges, the width of the distal penumbra can vary by a factor of 2.5. When the energy spread was controlled within the defined limits, the dose nonuniformity was less than $\pm 3\%$. To facilitate understanding of the results, the data were compared to the measured and Monte Carlo calculated data from a variable extraction energy synchrotron which has a narrow spectrum for all energies. Dose uniformity is only maintained within prescription limits when the energy spread is controlled. At low energies, a large spread can be beneficial for extending the energy range at which a single range modulator device can be used. An MLFC can be used as part of a feedback to provide specified energy spreads for different energies. © 2009 American Association of Physicists in Medicine.

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I. INTRODUCTION

Therapeutic proton beam delivery offers the ability to cover a target with a uniform dose while yielding sharp dose gradients at the periphery of the target. The energy spectrum of the proton beam is important for two reasons: The slope of the distal edge of the depth dose distribution and the design of the scheme to vary the penetration depth of protons into the patient. The modulation of penetration, often called range modulation, is used to cover a finite sized target with a desired dose distribution (usually uniform) while providing large dose gradients at the periphery of the target. This

modulation can be accomplished by delivering a sequence of energies, often referred to as energy layers. A major parameter in optimizing the range modulation scheme is the width of the peak of the nonmodulated depth dose distribution. Although the width of the peak, especially for ranges larger than 10 cm in water, is dominated by straggling due to discrete stochastic energy losses and multiple Coulombic scattering generated within the patient, the spectrum of proton energies entering the patient may also affect the peak width. For a given range, a narrow entrance energy spectrum results in a narrow peak that yields a steep distal edge dose gradient

FIG. 1. (a) A range modulation scheme for a small energy spread beam. (b) A range modulation scheme for a large energy spread beam. (c) Calculated dose distributions in water for small and large energy spread beams. Note that for clarity, complete modulation schemes are not given in (a) and (b).

but requires energy layers to be closely spaced so as to achieve a uniform dose distribution. Conversely, a wide energy spectrum allows the energy layers to be spaced further apart but yields a shallower distal edge dose gradient. Figure 1 illustrates the difference in modulation schemes and depth

dose distributions for beams with small and large incident energy spreads. The differences in the depth dose distributions are most apparent at the distal edge and at the proximal and distal ends of the modulated region. For beams with deeper penetration, the differences will be smaller. For beams with shallower penetration, the differences will be greater.

There are many items that can affect the spectrum of protons incident upon the patient surface and many methods available to perform range modulation. The first of these are the accelerator and energy selection systems. A synchrotron, being a cyclical machine, provides extracted beams with very small energy spreads. The full width at tenth maximum (FWTM) spread is typically less than 0.1% of the extracted energy.¹ This relative spread changes very little with the extraction energy—which may range from 17 to 260 MeV (Ref. 2) and is therefore negligible at all incident energies compared to the energy spread generated within the patient. A cyclotron, also being a cyclical machine, provides extracted beams with very small energy spreads. Typically the native spread is between 0.1% and 0.3%.³ The extraction energy of a cyclotron may be varied by changing its magnetic field or by accelerating doubly charged hydrogen and stripping the electrons at the orbit at which the desired energy is reached. For therapy purposes, however, most vendors have chosen to design their cyclotrons to extract protons at a single fixed energy and then use a variable thickness range shifter (RS) to change the penetration of the beam into the patient. This RS, however, causes straggling and different selected energies at the entrance of the patient will have different energy spreads. Table I gives both the full width at half maximum (FWHM) and FWTM energy spreads of a 250 MeV proton beam passing through three typical RS materials (water, carbon, and beryllium) for almost identical water-equivalent depths. The values were calculated using the Monte Carlo program MCNPX.⁴ Except for the largest thickness, the energy spreads are almost identical for the three materials.

If the RS is placed inside the nozzle, the energy spread that reaches the patient will always be large resulting in a large constant distal edge width, and the design of the modulation scheme will depend upon the different scattering of

TABLE I. Energy spread of 250 MeV protons through various materials.

Water $\rho=1.00 \text{ g/cm}^3$				Carbon $\rho=2.22 \text{ g/cm}^3$				Beryllium $\rho=1.82 \text{ g/cm}^3$			
Thickness (mm)	Peak energy (MeV)	FWHM (%)	FWTM (%)	Thickness (mm)	Peak energy (MeV)	FWHM (%)	FWTM (%)	Thickness (mm)	Peak energy (MeV)	FWHM (%)	FWTM (%)
0	250.0	0.038	0.075	0	250.0	0.038	0.075	0	250.0	0.038	0.075
50	230.1	0.79	1.45	24.69	230.5	0.80	1.44	33.29	230.4	0.77	1.42
100	208.9	1.27	2.33	49.33	209.7	1.25	2.32	66.58	209.5	1.25	2.32
150	186.3	1.78	3.28	74.07	187.5	1.79	3.26	99.87	187.1	1.72	3.28
200	161.2	2.59	4.73	98.77	163.7	2.50	4.63	133.16	162.9	2.58	4.68
250	134.1	3.78	6.94	123.46	136.7	3.73	6.85	166.44	135.9	3.79	7.04
300	101.7	6.27	11.8	148.15	105.5	6.18	11.3	199.73	103.9	6.32	11.8
350	58.4	17.1	33.2	172.84	64.7	14.6	27.9	233.02	62.5	15.6	30.6

FIG. 2. Schematic of the energy selection system in the fixed horizontal beamline at the Midwest Proton Radiotherapy Institute.

protons for different rangesifter thicknesses. If the RS is placed upstream of the nozzle, the entrance energy spectrum reaching the patient may be controlled by selecting a portion of the spectrum generated by the RS. The desired portion of the spectrum may be selected by passing the beam through a bending magnet and then inserting a collimator slit to reject those protons having unwanted energies. Figure 2 illustrates this process. It is obvious from Table I and Fig. 1 that trying to limit the energy spread to the small spread initially output from the accelerator would result in throwing away most of the protons. On the other hand, Fig. 1 shows that even a moderate energy spread of 3.5% produces a distinctly shallower distal edge gradient. Allowing a very large energy spectrum, such as the 11% to greater than 30% produced by large thicknesses of RS given in Table I, would thus result in unacceptably shallow distal edge gradients. In addition, the different energy protons would transit the beamline through different paths requiring larger magnets and vacuum beam pipes. Careful consideration of possible effects of beamline design on the entrance energy spectrum and depth dose distribution is thus desired.

A side issue concerning energy selection systems is whether the RS, magnets, and slits should be located immediately after the beam exit of the cyclotron or close to each treatment room. If the RS is placed near the accelerator, the spread of proton energies results in a large dispersion of paths through each downstream magnet. This RS placement requires many expensive large bore magnets. If the RS is placed near the treatment room, then the dispersion of paths occurs through only a few magnets and fewer large bore magnets are required saving costs. Downstream RSs also allow the main trunk line to operate at a single energy thereby stabilizing operations and improving beam switching time between treatment rooms. This ultimately improves the operating efficiency and the patient throughput. On the other hand, multiple energy selection systems are required if placement near the treatment room is desired possibly offsetting any cost gains of using smaller magnets. Additionally, the RS

system is a source of neutrons with the number of neutrons produced increasing with increasing RS thickness. Multiple RSs in the trunk line may require more distributed shielding compared to a single RS close to the accelerator, but the increase in capital cost may eventually be compensated by higher stability of operation and higher patient throughput.

In addition to the accelerator and energy selection systems, the range modulation method can affect the entrance energy spectrum. Range modulation can be performed by extracting different energies directly from the accelerator or through a RS and energy selection system but most often physical range modulation devices (RMDs) placed inside the nozzle have been used. In this case, two design approaches may be possible and they are described here.

If a physical RMD is placed upstream in a nozzle, some protons scattered by the thick part of the RMD are collimated within the nozzle and do not make it to the patient. Because the scattering angle is different for different energies, a different RMD may be required every 3–5 cm of range to cover the normal treatment energies, typically 70–250 MeV.⁵ Alternatively, a single (or few) RMD may be used and the beam current modulated to compensate for the scattered protons.⁶ This solution, however, requires that several diverse components of the delivery system be highly synchronized and a vigilant quality assurance program established to ensure that the desired modulation is always achieved.

In another design approach, where a physical RMD is placed downstream in a nozzle close to the patient, the RMD appears as if it is part of the patient and protons scattered from thick parts of the RMD make it to the patient. In this method, one RMD is adequate to cover a large range of energies without beam current modulation. Despite this apparent advantage, this method is not always implemented in practice because the RMDs are much larger typically requiring installation by hand.

Table II is a chart of several different beam delivery methods currently used or proposed for proton therapy. All of these methods should work satisfactorily as long as the spectrum entering the nozzle is adequately controlled to match the spectrum when the RMD was designed and commissioned. The Midwest Proton Radiotherapy Institute (MPRI), which uses method 3, is currently the only clinical facility that uses different energy slit settings for different energies. Before the beam enters the treatment room, an energy selection system with a multilayer Faraday cup (MLFC) inserted into the beamline upstream of the nozzle (see Fig. 2) is used to measure the entering proton energy spectrum. If the energy spread measured by the MLFC is outside a required tolerance, the beam delivery system (BDS) issues an error message to the accelerator operator, requiring an adjustment of the energy spectrum. Although the concept of using energy selection systems and multilayer Faraday cups in charged particle beams is not new,⁷ published details about the use of these devices and their calibration in clinical proton beams are sparse.^{8–10} This article reports on the calibration of the MLFC, the effect of energy selection slit setting on the range modulation scheme, dose uniformity, and distal

TABLE II. Beam delivery methods.

No.	Energy selection	Energy spread	Range modulation	Energy spread mitigation	Intensity loss
1	Synchrotron with variable extraction energy	Small for all energies	Downstream propellers	Ripple filter for low energies	None
2	Synchrotron with variable extraction energy	Small for all energies	Upstream propellers	Different propellers for different energy ranges combined with gating	Moderate
3	Cyclotron with upstream RS and slit	Increases with decreasing energy	Downstream propellers	Variable energy spread slit	Large
4	Cyclotron with upstream RS and slit	Increases with decreasing energy	Upstream propellers	Modulate beam intensity	Large
5	Cyclotron with nozzle RS	Constant for all ranges	Upstream propellers	None	Small
6	Linac with variable extraction energy	Small for all energies	Energy stacking	None	None
7	Laser with upstream slit	Increases with increasing energy	Multi leaf energy slit	Variable energy spread slit	Large

edge gradient under various conditions, and optimization of the energy slit settings at MPRI. Comparisons are also provided with measurements made for a system using beam delivery method 1.

II. METHODS AND MATERIALS

II.A. Description of beamline

Proton beams are provided to each of the treatment rooms at MPRI by the Indiana University Cyclotron Facility cyclotron.⁹ The energy of the protons from the cyclotron is 208.4 ± 0.2 MeV, as monitored continuously by an MLFC installed in a beam dump when the beam is not used for treatment or an experiment.¹¹ This energy corresponds to a range in water of 27.0 cm. Each treatment room is provided with its own rangeshifter that is comprised of a pair of beryllium wedges. The rangeshifter wedges adjust the range of the primary beam in depth steps of 0.02 cm water equivalence.¹² After passing through the wedges, next the beam passes through an energy analyzing magnet and adjustable slit (see Fig. 2). The momentum transport through this magnet without the slit is about 2.2%.¹² After the slit but before entering a nozzle, an MLFC may be inserted into the beamline to act as a surrogate for the water phantom for verifying the depth dose distribution to be delivered to the patient, specifically the range and width of the peak. This MLFC consists of 60 alternating layers of 0.5 mm aluminum conducting plates and 0.076 mm Kapton insulating plates allowing the position of the peak to be measured over a water-equivalent range of 7.20 cm with a resolution of 0.12 cm. Ranges between 7.2 and 37.5 cm of water can be measured by inserting into the beam one out of the seven copper blocks with various thicknesses mounted on a carousel just proximal to the MLFC according to the required beam range. For

simple operation, the difference in thickness between each block and the next thicker block is a constant water-equivalent thickness. Details of the beamline configuration between the trunk line and the fixed beamline room can be found elsewhere.¹³

At MPRI, the treatments are prescribed in range rather than energy. Before each treatment commences, the range and range spread measured with the MLFC are compared against predefined values, and treatments are not allowed if the disagreement is greater than ± 0.1 cm. Details of the MLFC calibration will be described in Sec. II B.

II.B. Calibration of MLFC for measuring range and energy spread

The output from the MLFC is the distribution of charge collected on the aluminum plates. For use as a surrogate, the MLFC was calibrated by relating the location of the peak of the charge distribution to the distal 80% dose (R_{80}) measured in a scanning water phantom for 13 beam energies at the treatment location with the scatterers removed from the nozzle. The peak location in the MLFC was determined by fitting a Gaussian shape to the distribution of charge in the MLFC channels. A small pedestal in the channels proximal to the peak of the MLFC distribution, as described by Gottschalk,¹⁴ does not introduce any significant error in obtaining the location and width of the peak. A modified linear fit to the data was used to determine the three calibration parameters in the equation below,

$$R_{80}(\text{cmH}_2\text{O}) = A_{\text{PT}} * \text{Peak}_{\text{MLFC}}(\text{Ch\#}) + (7 - \text{POS}) * A_{\text{BT}} + A_{\text{OS}}, \quad (1)$$

where A_{PT} is the water-equivalent thickness of a single MLFC layer, $\text{Peak}_{\text{MLFC}}(\text{Ch\#})$ is the location of the peak in

terms of the channel number, POS is an integer that designates which copper block is inserted in front of the MLFC, A_{BT} is the constant difference in water-equivalent thickness between blocks, and A_{OS} is an offset value. The parameter values found from the fitting process were 0.118 ± 0.0007 cm for A_{PT} , 4.831 ± 0.016 cm for A_{BT} , and 1.841 ± 0.023 cm for the offset A_{OS} .

In addition to calibrating the MLFC for use as a range measurement surrogate, the MLFC was also calibrated for use as a surrogate for the peak width in water. For convenience, the width of the peak between the proximal and distal 50% doses (W_{50}) measured in water at the treatment location was related to the measured FWHM width of the peak in the MLFC. Here again, matched MLFC and water phantom measurements were used, this time with Eq. (2), to determine two calibration parameters,

$$W_{50}(\text{cm}_{\text{H}_2\text{O}}) = B_1 * \text{FWHM}_{\text{MLFC}}(\text{Ch \#}) - B_{OS}, \quad (2)$$

where B_1 is a conversion factor, $\text{FWHM}_{\text{MLFC}}(\text{Ch \#})$ is the measured full width at half maximum of the MLFC charge distribution expressed in units of channel numbers, and B_{OS} is an offset. The parameter values found from the fitting process were 0.4035 ± 0.02 cm for B_1 and 0.3177 ± 0.12 cm for B_{OS} .

Before each treatment, both the location and width of the peak of a tuning beam within the MLFC are measured, converted to the appropriate range (R_{BDS}) and energy spread (ES_{BDS}) surrogates for the beam delivery system using Eqs. (3) and (4), and verified to match the values prescribed for the treatment,

$$R_{\text{BDS}}(\text{cm}_{\text{H}_2\text{O}}) = A_{PT} * \text{Peak}_{\text{MLFC}}(\text{Ch \#}) + (7 - \text{POS}) * A_{BT} + A_{OS}, \quad (3)$$

$$\text{ES}_{\text{BDS}}(\text{cm}_{\text{H}_2\text{O}}) = B_1 * \text{FWHM}_{\text{MLFC}}(\text{Ch \#}) - B_{OS}. \quad (4)$$

It is important to note that Eqs. (3) and (4) apply only to the range and peak width measured when the dual scatterers (see Sec. II C) are removed from the nozzle. For use as surrogates, the ranges defined for a specific patient plan, given by the distal 90% depth dose with all devices in the beamline, are converted to R_{BDS} parameters when the prescribed range and modulation width are exported from the treatment planning system.

II.C. Description of nozzle

The fixed horizontal beamline at MPRI uses a dual scatterer system to form large uniform proton fields for treatment. Details of these beamline devices were reported in an earlier publication.¹⁵ Two snouts (also known as cones or applicators) are used in this beamline to provide maximum field sizes projected to the isocenter (treatment location): a 10 cm diameter, the so-called “small field snout,” and a 20 cm diameter, the so-called “large field snout.” Each of these snouts is provided with a single contoured second scatterer and two uniform thickness first scatterers. One of the first scatterers for each snout is used for low energies with

beam ranges between R_{BDS} 6 and 14 cm, while the other first scatterer is used to cover high energies with beam ranges between R_{BDS} 14 and 27 cm. To simplify the exchange of information between treatment planning and treatment delivery, the first scatterers were designed to maintain the range loss through the nozzle for a given snout (i.e., second scatterer) to the same value regardless of the first scatterer and beam range applied. This was achieved by carefully combining polycarbonate with lead during the fabrication of the low-energy first scatterer. The resulting range losses for the small and large field snouts using these combinations of scatterers are 1.1 and 2.5 cm, respectively, regardless of the range in the patient. An additional range difference that must be accounted for when calibrating the MLFC is the difference between the R_{90} of a modulated beam and the R_{80} of a nonmodulated beam; these differences are 0.1 and 0.05 cm for the high- and low-energy beam ranges, respectively. Once the scatterers were designed and inserted into the nozzle, the effect of various slit settings on the R_{80} was investigated for various R_{BDS} values.

II.D. Validating modulator design for different ranges and energy spreads

A goal of the MPRI facility was to use a single range modulator device for all target volumes having approximately the same dimension in the depth direction over a large range of beam ranges (i.e., penetration depths). The uniformity of the depth dose profile over the target region for a given range modulator device can vary in its smoothness and slope with beam range if the widths of the individual peaks vary largely with the beam range. To characterize the width of the peak as a function of R_{BDS} and slit setting, depth dose curves were measured with a Markus ionization chamber (PTW type 23343) in a scanning water phantom using a large area parallel plate transmission ion chamber inserted into the nozzle just upstream of the range modulator as a reference beam intensity detector.

Modulator devices were designed assuming that the peak width was identical for various beam ranges. The measured base depth dose distribution used for the design was taken from a scan of a beam with an R_{BDS} of 17 cm range and a W_{50} of 2.0 cm, which corresponded to the slit setting of 0.4. Figure 3 shows these base depth dose data. Also shown in Fig. 3 are the same depth dose data but shifted in depth to shallower ranges and scaled in dose by a weighting factor. Each range shifted depth dose curve represents a layer of modulation. The modulator device design process optimizes the interlayer step sizes and layer weighting factors. In the optimization process used here, all step sizes between layers were set identical. For an 8.0 cm modulation, which is shown here as an example, a total of 13 layers was used. A cost function was used to optimize the weights of each layer. The cost function minimizes the differences between a reference dose and the calculated doses at all depths between $R_{80} - M_{\text{max}}$ and $R_{80} - W_d$, where M_{max} is the maximum thickness (in units of water-equivalent thickness) of the modulator device and W_d is the difference in depths of the distal 20%

FIG. 3. Modulator design with the base depth dose distribution of a non-modulated beam shown as a solid curve. The top curve (open circles) is a depth dose distribution for a nominal 8 cm modulation calculated by summing weighted nonmodulated layers shown by dashed curves. The weight of each layer was obtained by minimizing a cost function that was calculated by summing the squares of differences between referenced and calculated doses at depths from $R_{80}-M_{\max}$ to $R_{80}-W_d$. W_d is the distance between the depths of the distal 20% and 80% dose levels and $M_{\max}$ is the maximum thickness (in water-equivalent thickness) of the modulator.

and 80% dose levels of the deepest penetrating layer. The modulation width, W_{SOBP} , for a manufactured modulator is defined as the difference in depth between the distal and the proximal 90% doses. The measured W_{SOBP} of a manufactured modulator may be a few millimeters different than the designed nominal width; however, $M_{\max}$ is always smaller than W_{SOBP} as seen in Fig. 3. The Appendix describes the relationship between W_{SOBP} and $M_{\max}$ for various conditions in more detail.

Once each modulator was manufactured according to its design, depth dose distributions were measured for modulated beams at various beam ranges and slit settings to determine if the uniformity was within $\pm 3\%$ at all depths within the desired modulated region. Depth doses for modulated beams were measured using the same equipment as for the nonmodulated beams.

With a spinning modulator, the beam penetration changes rapidly. If the ion chamber is scanned simultaneously with the range changes, large fluctuations are observed at depths close to the distal edge as shown by the thin line in Fig. 4 for a beam with an R_{BDS} of 10 cm, modulation width of 8 cm, and a slit setting of 0.3. These fluctuations are the result of no dose being delivered to the ionization chamber when the thick part of the modulator is between the beam source and the chamber. A smoothing algorithm based upon a least-squares approach was applied to the measured depth doses before assessing the dose uniformity. Thirty-one neighboring depths with 0.1 mm resolution were used for the data smoothing. The corresponding smoothed depth dose is shown as the thick line in Fig. 4. Incidentally, Fig. 4 demonstrates that for the chosen interlayer step sizes, a slit setting

FIG. 4. Depth dose curve for a beam with a R_{BDS} of 10 cm, slit setting of 0.3, and 8.0 cm modulation width. The thin line is the data acquired with a parallel plate ionization chamber in a water phantom using a continuous scan mode. The thick line shows the same data after application of smoothing with a least-squares approach. The solid circles show a depth dose measured with the MLIC.

of 0.3 at this beam range is too restrictive of the entrance energy spectrum to obtain a uniform treatment region. This aspect will be discussed further in Sec. III.

To reduce the beam time required for data acquisition of range modulated beams, and also to avoid large fluctuations in depth dose measurements, some depth dose curves were measured with a multilayer ionization chamber (MLIC).¹⁶ The MLIC allows the simultaneous measurement of dose at 125 depths with a spatial resolution of 0.182 cm and a 1.0 s recycle integration time during data acquisition. This was particularly useful for investigating the effect of different slit settings at different ranges. In addition to the water phantom scans, Fig. 4 also shows the MLIC measured depth dose using the same beam conditions demonstrating the equivalence of the two measurement techniques.

III. RESULTS AND DISCUSSION

III.A. Range and energy spread of nonmodulated beams

Figure 5 shows the MLFC charge distributions of non-modulated proton beams for three beam ranges each acquired with three different slit settings. A setting of “1.0” indicates that the slit is fully open, while a setting of “0.0” indicates that the slit is completely closed. The beam current in the trunk line was constant for all measurements at a given beam range. For each beam range, the measured data were normalized to the peak value with a slit setting of 1.0. The MLFC peak widths were calculated using Eq. (4) (ES_{BDS}). The MLFC peak locations, heights, and widths are given in Table III. Table III also includes data for an additional beam range not shown in Fig. 5. The R_{BDS} beam ranges derived from the measured peak locations according to Eq. (3) are 10.01, 16.03, and 20.05 cm with MLFC block positions of 6, 5, and 4 for the requested R_{BDS} ranges of 10.0, 16.0, and 20.0 cm,

FIG. 5. Measured MLFC charge distributions for R_{BDS} ranges of 10, 16, and 20 cm. For each R_{BDS} , the measured data are normalized to the peak value when a slit setting of 1.0 was used. The MLFC block index positions were at 6, 5, and 4 for ranges of 10, 16, and 20 cm, respectively. Open circles with a solid curve represent data when the slit was fully open (1.0). Solid squares with a dashed curve are for a slit setting of 0.5. Solid circles with dot-dashed curves are for a slit setting of 0.3. All curves represent Gaussian fits to the measured data.

respectively. These results show good agreement between the requested and the delivered beam ranges. As can be seen from Table III, for all slit settings for each given R_{BDS} , the MLFC peak location and the location of the R_{80} in water remain constant, suggesting that the range of the proton beam did not depend on the slit setting. For R_{BDS} ranges of 20, 16, 10, and 6 cm, the width of the peak decreased by approximately 0, 1, 2, and 3 channel units, respectively, between slit settings of 1.0 and 0.3. For all three higher ranges, the height of the Peak_{MLFC} varied only by about 1% when changing the slit setting from 1.0 to 0.5 but decreased by 4%, 15%, and 21% for R_{BDS} of 20, 16, and 10 cm respectively, when the slit setting was reduced to 0.3. The significance of these results is that the slit does provide energy spectrum control for R_{BDS} less than 20 cm; however, at these ranges, the energy spectrum control comes with the cost of reducing the intensity of the desired energy protons that reach the nozzle. For R_{BDS} above 20 cm, the slit provides little control over the entrance energy spectrum due to the fact that the beam size at these energies is smaller than the slit opening for most of the slit settings used in this work.

Using beams with the same R_{BDS} ranges and slit settings as used for the plots shown in Fig. 5, depth dose curves were measured with a water phantom placed in the treatment room with the small snout and its associated scatterers inserted. A 10 cm circular aperture was used. The measured depth dose curves are plotted in Fig. 6. The measured R_{80} depths were 8.9, 14.9, and 18.9 cm for R_{BDS} ranges of 10, 16, and 20 cm, respectively. For both first scatterers used with the small snout and all slit settings, a constant 1.1 cm range shift was seen for all R_{BDS} . For each R_{BDS} , the depth of R_{80} was found to be invariant as a function of slit setting, as was found with

TABLE III. Results from nonmodulated beam measurements using MLFC and ionization chamber in water phantom. Column 3–5 give the location, height, and FWHM of the peaks extracted from MLFC charge distributions. Column 6 gives the ES_{BDS} in units of cm of water according to Eq. (4). Column 7 gives the W_{50} measured with the water phantom at the treatment location.

R_{BDS} (cm)	Slit opening	Peak location (Ch #)	Peak height (Norm*)	FWHM (Ch #)	ES_{BDS} (cm _{H₂O})	W_{50} (cm _{H₂O})
20	0.3	32.5	0.96	6.06	2.13	2.19
	0.5	32.5	0.99	6.55	2.33	2.32
	1	32.5	1.00*	6.56	2.33	2.32
16	0.3	39.4	0.85	5.21	1.78	1.84
	0.5	39.4	0.99	6.31	2.23	2.21
	1	39.4	1.00*	6.38	2.26	2.27
10	0.3	29.35	0.77	3.49	1.09	1.28
	0.5	29.35	0.99	5.12	1.75	1.81
	1	29.35	1.00*	6.05	2.12	2.03
6	0.3	—	—	2.37	0.64	—
	0.4	—	—	2.93	0.87	—
	0.5	—	—	3.66	1.16	—
	0.6	—	—	4.35	1.44	—
	0.7	—	—	4.88	1.65	—
	1	—	—	5.32	1.83	—

FIG. 6. Depth dose curves measured in a scanning water phantom for R_{BDS} ranges of 10, 16, and 20 cm. Measurements were obtained with a parallel plate ionization chamber using a continuous scanning mode. A 10 cm circular aperture was used with the scatterers of small snout placed inside the beamline. Open circles represent data for the energy selection slit wide open (1.0), solid curves represent data for the slit set to 0.5, and dashed curves represent data for the slit set to 0.3.

the $Peak_{MLFC}$ for the MLFC measurements. Peak widths extracted from the measured depth dose distributions in water (W_{50}) are given in column 7 of Table III in units of centimeters of water. The variation in W_{50} , listed in Table III, showed a similar pattern as the ES_{BDS} measured with the MLFC. These results demonstrate that the MLFC measurements can be used to verify the range and energy spread of the treatment beam even when the scatterers and snout are in place. The MLFC surrogate values of R_{BDS} and ES_{BDS} can predict the R_{80} measured in water within 0.05 cm and the W_{50} peak width in water within 0.2 cm.

III.B. Effect of energy spread on modulated beam depth dose distributions

Figures 7(a)–7(c) show depth dose curves for a nominal 8 cm range modulation and beam ranges of 20, 16, and 10 cm, respectively, each with three energy slit settings. For the beam with a R_{BDS} of 20 cm [Fig. 7(a)], the effect of different slit settings on the dose uniformity within the target region is small. This is expected because the W_{50} is only slightly different (<0.1 cm out of 2 cm) for different slit settings. For the beam with a range of 16 cm [Fig. 7(b)] and a slit setting of 0.3, a ripple pattern ($\pm 3\%$) appears close to the distal edge. With this slit setting, the W_{50} is 0.22 cm smaller than when the base depth dose distribution was acquired and which was used for the range modulator design. For the beam with a R_{BDS} of 10 cm [Fig. 7(c)], a large ripple pattern (up to $\pm 5\%$) was seen for a slit setting of 0.3 but very little ripple was seen for a slit setting of 1.0. The W_{50} for the slit setting with the large ripple was almost half of what was used for the range modulation optimization.

Figure 8 shows depth dose distributions measured for a range of 6.0 cm and nominal 8.0 cm modulation with several slit settings. A very large ripple pattern ($\pm 21\%$) was ob-

FIG. 7. Measured depth dose curves of nominal 8 cm modulation for R_{BDS} of 10, 16, and 20 cm with three different slit settings. Solid curves were obtained from scans in water and application of least-squares smoothing. Points represent measurements with the MLIC. The data with slit settings of 0.5 and 1.0 have been offset by -20% and -40% , respectively, for ease of viewing.

FIG. 8. Modulated depth dose curves for a R_{BDS} of 6.0 cm and nominal 8.0 cm modulation with slit settings of 0.3, 0.4, 0.5, 0.6, and 1.0. Data were offset by -20% , -20% , and -40% for slit settings of 0.5, 0.6, and 1.0, respectively, to ease viewing. A large ripple up to $\pm 21\%$ was seen for the 0.3 slit setting.

FIG. 9. Depth dose peak widths (W_{50}) as a function of range. Open squares represent Monte Carlo widths calculated in a water phantom. Solid squares represent measured widths from LLUPTF beamlines. Solid circles represent measured widths from the MPRI fixed horizontal beamline with a slit setting of 1.0 (fully open).

served when the slit was set to 0.3 and, in fact, the ripple pattern extended completely through the target region from the distal to the proximal edge. Even with the largest slit setting of 1.0, however, a small but acceptable ripple pattern ($\pm 2\%$) was observed.

III.C. Limitations on energy spread control

For an accelerator that extracts beam at multiple energies, the entrance energy spectrum will be small for all ranges, and the width of the peak measured in water at the treatment location will decrease as the range decreases. For an accelerator that extracts beam at a single fixed energy and produces different range beams by use of a RS, the entrance energy spectrum increases as the range of the beam decreases. If the entire energy spread generated by straggling in the RS is transported to the treatment location, then the width of the peak measured in water should be constant as a function of range.

Figure 9 shows plots of W_{50} as a function of range in water. Open squares represent simulations using the Monte Carlo program MCNPX.⁴ For the simulation, protons were incident upon a water phantom with an approximately Gaussian spread of energies having a full width at tenth maximum 0.1% of the nominal entrance energy for all ranges.¹ Solid squares represent W_{50} measured in water or polystyrene for several beams from the variable extraction energy synchrotron at the Loma Linda University Proton Treatment Facility (LLUPTF). Except for small ranges where the energy spectrum from the synchrotron has not been measured and therefore possibly not modeled correctly, fairly good agreement was seen between the MCNPX water simulated and LLUPTF beamline measured W_{50} . Solid circles represent measurements of W_{50} for the MPRI fixed horizontal beamline with the energy slit fully open. For the maximum beam range at MPRI of 27 cm, i.e., when no RS is inserted, W_{50} is 2.3 cm

and closely matches the LLUPTF and MCNPX values as expected. Theoretically, all MPRI W_{50} with the slit fully open should have this same value. The graph shows, however, that as the range decreases below 16 cm, W_{50} also decreases suggesting that the beamline between the RS and the isocenter does not transmit the full energy spectrum generated by the proton beam passing through the RS.

Section II D stated that a slit setting of 0.4 and a R_{BDS} of 17.0 were selected for measurement of the base depth dose distribution. This setting provided a W_{50} value of 2.0 cm which is near the middle of all possible W_{50} values for all possible delivered beam ranges between 6.0 and 27.0 cm. The selected R_{BDS} was likewise selected near the center of all possible ranges. Selecting a W_{50} and R_{BDS} near the center of all possible values provides a better chance for designing a modulator device that can be used for all situations. One feature of the MPRI nozzle is that the modulator device is located relatively close to the patient. This allows almost all of the energy spread generated by the modulator to be transmitted to the patient. This means that the W_{50} of each energy layer, for a given maximum range, will be identical to the base layer W_{50} . If the W_{50} is close to the W_{50} of the base depth dose distribution used for designing the modulator device, then the ripple in the dose distribution over the target region will be minimized. Energy spectrum control via slit setting should afford the ability to maintain the W_{50} close to the W_{50} of the base depth dose distribution and ensure uniform depth dose distributions.

Another concern about using a single modulator device to produce a fixed modulation width over a wide range of beam ranges was that the dose distribution might be sloped over the target region; for example, too much dose near the proximal end of the target and too little dose near the distal end of the target. Careful study of Figs. 7 and 8 reveals no discernible slope for all ranges and all slit settings. Generating a slope of the dose distribution for a modulated proton beam requires a variation in the weighting factors of the energy layers. This would be caused not so much by changing the width of the peak but by producing different magnitudes of scatter inside the nozzle. As stated earlier, the nozzle with a downstream modulator device transmits most of the scattered protons and reduces proton losses between the modulator and the patient. As a result, the variation in slope on modulated proton beam is minimal even when large ripple is seen for low beam ranges with small slit settings.

In addition to influencing the uniformity of dose over the target region, the entrance energy spectrum can affect the dose gradient at the distal edge of the modulated region. A larger energy spread entering the nozzle produces a shallower distal dose gradient. The parameter W_d , introduced in Sec. II D, can be used to present and analyze the dose gradient. A small W_d is preferred when avoiding normal structures distal to the target, whereas a wider W_d is sometimes desired for matching or patching multiple fields. Figure 10 plots W_d as a function of R_{BDS} for three different slit settings for the MPRI beamline and for three different nozzle types for the LLUPTF. Triangles represent LLUPTF values for a nozzle with a double scattering system designed for field

FIG. 10. Distal dose gradient (W_d) versus R_{BDS} for various beam conditions at MPRI and LLUPTF. Detailed descriptions of each beam are given in the text. Curves/lines are plotted for each group of matching beams to guide the eye.

sizes of 18–22 cm in diameter. A line has been fit to the points between R_{BDS} of 16 and 38 cm to help guide the eye. The W_d for a LLUPTF radiosurgery beam is represented by a solid square. This beam, which employs a single scattering system, is only used for field sizes less than 4 cm in diameter and R_{BDS} less than 12 cm. For these small R_{BDS} , the W_{50} is too small for the standard modulator devices designed at a higher energy and therefore a one step ripple filter is used in conjunction with the standard modulator device to produce uniform dose over the target region. This technique slightly increases both the entrance energy spread and the W_d . The point represented by a solid diamond is for the LLUPTF eye beamline that uses a single scatterer system. This beamline typically uses protons with energy of 100 MeV and a RS within the nozzle to decrease the range from 7.8 to 3.6 cm. This technique also slightly increases both the entrance energy spread and the W_d compared to a nonrangeshifted beam with the same range in the patient. For the MPRI data, different trends are seen for different slit settings. When the slit is fully open (open circles), the W_d values remain fairly constant as R_{BDS} decreases. When the slit is set to 0.3 (open squares), the W_d decreases with decreasing R_{BDS} in a fashion similar to that seen for the non-RS LLUPTF beams. The maximum difference in W_d for different slit settings was found for small ranges resulting in the width increasing by a factor of 2.5. Some proton therapy cyclotrons operate at higher energies than at MPRI, between 230 and 250 MeV, and therefore the range must be shifted to a much larger extent. The difference between W_d for different energy spectra (slit settings) for these machines would be much larger.

The initial premise for the modulator design optimization at MPRI was that a constant W_{50} was required to use a single modulator device for all beam ranges. It appears from Figs. 7 and 8, however, that for a given range modulation width, the ripple can be kept within acceptable tolerances (within $\pm 3\%$) using a single modulator device and a fully opened

FIG. 11. Clinical slit setting scheme and the resulting entrance range spread (i.e., energy spectrum). The measured W_{50} widths as a function of requested R_{BDS} are plotted as open circles. The corresponding energy slit settings are plotted as solid squares. A constant W_{50} of 2.0 cm was achieved for R_{BDS} between 10 and 20 cm.

slit for all R_{BDS} between 6 and 27 cm even though W_{50} varies from 1.1 to 2.4 cm. As seen in Fig. 6 and Table III, the W_{50} is relatively insensitive to the slit setting for R_{BDS} greater than 20 cm. This is because of two phenomena. First, the beam diameter is so small exiting the RS that the slit has little effect. Second, the width of the peak in the phantom at the treatment location is dominated by straggling in the phantom rather than in the RS. For these high beam ranges, therefore, the slit is set to its fully open position for clinical treatments. For R_{BDS} between 10 and 20 cm, a constant W_{50} of 2.0 cm can be achieved, as shown in Fig. 11, by varying the slit setting between 0.6 and 0.2. This scheme was used during the commissioning of the beamline for patient treatments; however, setting the slits fully open may be preferable. As can be seen from Table III and Fig. 9, for R_{BDS} ranges of 10 cm and below, the peak width measured by the MLFC is equal or less than the expected value of W_{50} and closing the slit for these beam ranges would only make the W_{50} narrower than the desired 2.0 cm while also decreasing the beam intensity. Below an R_{BDS} of 10 cm, depth dose curves using modulators designed at higher energy become nonuniform at MPRI for small slit settings (see Figs. 7 and 8). This is similar to what is seen at LLUPTF where no rangeshifters or energy slits are employed and there is little energy spread entering the nozzle. At LLUPTF, to control the energy spread somewhat, a ripple filter is used in conjunction with the standard modulator device to generate a second, slightly lower energy thus increasing the energy spread and making the dose distribution more uniform. At MPRI, the energy spread can be maintained simply by setting the energy slit fully open to transport the energy spread generated by the rangeshifter.

IV. CONCLUSIONS

The energy spread in the beam transport system at MPRI has been studied with respect to its effect on range, modula-

tion width, distal penumbra, and uniformity of the dose distribution over the modulated region. The RS method of producing beams of different ranges, together with a large energy spread acceptance, results in a smaller change in the nonmodulated peak width of a depth dose distribution than does the method where the different ranges are produced by direct extraction of different energies from the accelerator. A control scheme has been devised that minimizes changes in the width of the peak of a nonmodulated depth dose distribution for some ranges. Nevertheless, for optimization of the modulator design, it is important to carefully select the base conditions for the measurement of the nonmodulated depth dose distribution. This will allow a modulator device designed for a fixed modulation width and a given beam range to be used over a wide range of beam ranges with acceptable nonuniformities. Even without using the devised energy spread control scheme, acceptable uniform dose distributions may be obtained for ranges between 10 and 20 cm because of the intrinsic acceptance of the beam transport system; however, better uniformity is obtained using the scheme.

For all clinically used modulators, the modulation width W_{SOBP} was found constant (within 3 mm) for all beam ranges and slit settings used in this work. The beam range was also found to be independent of the slit setting for all R_{BDS} studied. The dose gradient at the distal edge of the modulated region with this RS beam and fully opened slit was found to be nearly constant with the beam range. Verification of the range and energy spread of the beam before each treatment is an important step to ensure that the patient receives the correct depth dose distribution and can be performed using an upstream surrogate device such as an MLFC.

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APPENDIX: RANGE MODULATOR DEVICES

Using the range modulation technique at MPRI, a separate modulation device is required for each desired modulation width. However, most treatment ranges for a given modulation width can be treated with a single device. During the commissioning of the beamline and treatment planning system, a nominal name was given to each modulator to indicate the nominal modulation width (W_{SOBP}). In reality, the measured W_{SOBP} of each modulator can vary from its nominal width for the design range by up to 3 mm. The variation in the measured W_{SOBP} for different beam ranges can also vary up to 3 mm. Table IV is a list of modulators manufactured for the MPRI fixed beamline. Column 1 is the nominal designed W_{SOBP} , column 2 is the measured W_{SOBP} with an R_{BDS} of 27 cm, and column 3 is the maximum thickness M_{max} of the device in terms of water equivalence. The

TABLE IV. List of manufactured modulators at MPRI. Column 1 is nominal width, column 2 is the measured width for R_{BDS} 27 cm, and column 3 is the maximum thickness (water equivalent) of the physical modulator.

W_{SOBP} (cm)		Modulator maximum thick (cm water)
Nominal	Measured	
2.6	2.59	2.16
3.3	3.19	2.79
4.1	3.98	3.51
5.1	4.97	4.43
6.0	5.92	5.18
7.1	6.90	6.01
8.0	7.84	6.86
9.0	8.65	7.52
10.0	9.85	8.53
11.0	10.58	9.26
12.3	11.96	10.21
14.0	13.90	11.57
16.4	16.20	13.79

difference between the modulation width and the maximal thickness increases with modulation width, as shown in Table IV and Fig. 12.

The nonlinear relationship between the modulation width and maximal thickness can be well modeled by a second order polynomial with values of 0.1186, 1.0838, and 0.0069 for the zero, first, and second order fit coefficients, respectively. This nonlinear relationship holds for all beam ranges because the W_{SOBP} does not vary much for different beam ranges. To demonstrate the generality of this relationship, measured W_{SOBP} with the corresponding M_{max} from the LLUPTF are also plotted in Fig. 12. Although the design criteria and approach are somewhat different at the two facilities, for example, the shoulders of the dose distribution at the proximal and distal ends of the modulated region are

FIG. 12. Nonlinear relationship between the measured W_{SOBP} and M_{max} . Closed circles are data from MPRI and open squares are data from LLUPTF. Solid curves represent second order polynomial fits for each set of data. A dashed line is provided as a reference for a linear condition.

higher at LLUPTF, a similar second order polynomial relationship can still describe the LLUPTF data with the zero, first, and second order fit coefficients being 0.0254, 0.944, and 0.0319.

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