An evaluation of spatial resolution of a prototype proton CT scanner

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Purpose: To evaluate the spatial resolution of proton CT using both a prototype proton CT scanner and Monte Carlo simulations.

Methods: A custom cylindrical edge phantom containing twelve tissue-equivalent inserts with four different compositions at varying radial displacements from the axis of rotation was developed for measuring the modulation transfer function (MTF) of a prototype proton CT scanner. Two scans of the phantom, centered on the axis of rotation, were obtained with a 200 MeV, low-intensity proton beam: one scan with steps of 4°, and one scan with the phantom continuously rotating. In addition, Monte Carlo simulations of the phantom scan were performed using scanners idealized to various degrees. The data were reconstructed using an iterative projection method with added total variation superiorization based on individual proton histories. Edge spread functions in the radial and azimuthal directions were obtained using the oversampling technique. These were then used to obtain the modulation transfer functions. The spatial resolution was defined by the 10% value of the modulation transfer function (MTF10%) in units of line pairs per centimeter (lp/cm). Data from the simulations were used to better understand the contributions of multiple Coulomb scattering in the phantom and the scanner hardware, as well as the effect of discretization of proton location.

Results: The radial spatial resolution of the prototype proton CT scanner depends on the total path length, \( W \), of the proton in the phantom, whereas the azimuthal spatial resolution depends both on \( W \) and the position, \( u_- \), at which the most-likely path uncertainty is evaluated along the path. For protons contributing to radial spatial resolution, \( W \) varies with the radial position of the edge, whereas for protons contributing to azimuthal spatial resolution, \( W \) is approximately constant. For a pixel size of 0.625 mm, the radial spatial resolution of the image reconstructed from the fully idealized simulation data ranged between 6.31±0.36 lp/cm for \( W = 197 \) mm i.e., close to the center of the phantom, and 13.79±0.36 lp/cm for \( W = 97 \) mm, near the periphery of the phantom. The azimuthal spatial resolution ranged from 6.99±0.23 lp/cm at \( u_- = 75 \) mm (near the center) to 11.20±0.26 lp/cm at \( u_- = 20 \) mm (near the periphery). Multiple Coulomb scattering limits the radial spatial resolution for path lengths greater than approximately 130 mm, and the azimuthal spatial resolution for positions of evaluation greater than approximately 40 mm for \( W = 199 \) mm. The radial spatial resolution of the image reconstructed from data from the 4° stepped experimental scan ranged from 5.11±0.61 lp/cm for \( W = 197 \) mm to 8.58±0.50 lp/cm for \( W = 97 \) mm. In the azimuthal direction, the spatial resolution ranged from 5.37±0.40 lp/cm at \( u_- = 75 \) mm to 7.27±0.39 lp/cm at \( u_- = 20 \) mm. The continuous scan achieved the same spatial resolution as that of the stepped scan.
Conclusions: Multiple Coulomb scattering in the phantom is the limiting physical factor of the achievable spatial resolution of proton CT; additional loss of spatial resolution in the prototype system is associated with scattering in the proton tracking system and inadequacies of the proton path estimate used in the iterative reconstruction algorithm. Improvement in spatial resolution may be achievable by improving the most likely path estimate by incorporating information about high and low density materials, and by minimizing multiple Coulomb scattering in the proton tracking system. © 2016 American Association of Physicists in Medicine. [http://dx.doi.org/10.1118/1.4966028]

Key words: proton imaging, computed tomography, spatial resolution, modulation transfer function, oversampling method

1. INTRODUCTION

The initial discussion of proton imaging versus x-ray imaging in the late 1970s pointed to the lack of spatial resolution in proton imaging compared with the already-successful x-ray CT technology. This led to the abandonment of proton CT as a diagnostic imaging modality. The recent increase in the number of proton therapy facilities, and the lack of imaging support for proton therapy in the treatment room, has led to a renewed interest in proton radiography and CT for improved range definition and treatment verification.1–3

The present procedure for proton therapy planning involves converting the Hounsfield value of each voxel in x-ray CT planning scans of the patient into proton stopping power via a stoichiometrically acquired calibration curve. However, since there is no unique relationship between Hounsfield values and proton stopping power, this procedure has inherent uncertainties of a few percent in the proton range, requiring additional distal uncertainty margins in proton treatment plans. Cone beam x-ray CT is now becoming available for image guidance and treatment plan verification; however, it has distinct disadvantages due to reconstruction artifacts and even larger range uncertainties.

In contrast to x-ray CT, proton CT measures the relative stopping power (RSP) with respect to water of the object directly, eliminating the need for Hounsfield-value-to-RSP conversion. In the prototype proton CT scanner that we have developed in recent years,4–6 a low-intensity energetic beam of protons traverses the phantom, entirely, and stops in a downstream energy/range detector. The entry and exit vectors of each proton are measured in order to determine a most-likely path estimate used in the iterative reconstruction algorithm. Improvement in spatial resolution may be achievable by improving the most likely path estimate by incorporating information about high and low density materials, and by minimizing multiple Coulomb scattering in the proton tracking system.

The spatial resolution of proton CT images is fundamentally limited by multiple Coulomb scattering (MCS), which determines the uncertainty of the MLP prediction. Due to MCS, straight-line projections are not accurate enough for clinically acceptable spatial resolution. For this reason, image reconstruction is better accomplished by using a MLP formalism.10,11 In its current form, the MLP formalism applies a Gaussian approximation of MCS in order to estimate the MLP of the proton through the object, assuming it is uniform and consists of water, which is an approximation. In addition to the MLP estimate, the formalism provides an uncertainty envelope for given entry and exit vectors of each proton. This uncertainty envelope provides a theoretical approximation of the limit of spatial resolution imposed by MCS. It is important to note that the primary purpose of proton CT is to create an accurate map of RSP for proton radiation therapy planning. As such, spatial resolution is intimately related to the accurate prediction of proton range when protons pass along high-contrast tissue interfaces, i.e., the prediction of range dilution effects.12

The modulation transfer function (MTF) is widely used to characterize the spatial resolution of imaging systems. The most common way to measure the MTF is to use a phantom with a thin, high-density metallic wire embedded orthogonally to the scanning plane. The reconstructed image of the wire yields the point spread function (PSF), while the Fourier transform of the PSF yields the MTF. However, in noisy or low-contrast images, the use of an edge phantom is the preferred alternative.13 In this method, blocks of material with high contrast, compared with the background, and sharp straight edges are used as test objects. The edge spread function (ESF) is obtained by overlaying many edge profiles, and is then differentiated to obtain a line spread function (LSF). The LSF can then be Fourier-transformed to obtain the MTF.

The acceptability of any proton imaging system depends on the relative importance assigned to visual quality of an image (spatial resolution and noise) and RSP accuracy.1 A theoretical study of spatial resolution in proton and carbon radiography using Monte Carlo simulations was performed by Seco et al.,14 and theoretical estimates of spatial resolution in proton CT have been published by Schneider et al. and Hansen et al.,9,15 but, to the authors’ knowledge, no comprehensive study of spatial resolution of an experimental proton CT scanner has been published. In this paper, we present the radial and azimuthal MTFs for a prototype proton CT scanner to characterize the spatial resolution that can be achieved with such a system using 200 MeV protons.

2. MATERIALS AND METHODS

2.A. Prototype proton CT scanner

The prototype proton CT scanner consists of a particle tracker composed of silicon strip detectors (SSDs) and a
multistage scintillator (MSS) for WEPL measurement. The particle tracker is composed of two particle telescopes, one upstream and one downstream from the phantom, each arranged into four layers of four 400 μm-thick silicon wafers with a strip pitch of 228 μm. Each layer of the silicon tracker measures the position of the protons with greater than 99% efficiency. The front telescope measures the coordinates and the angle of the proton before it enters the phantom, whereas the rear telescope measures the coordinates and the angle of the proton after it exits the phantom. The telescopes have a total sensitive area of 8.6×34.9 cm². The tracking planes interface through custom readout ICs and a high-speed data acquisition system based on field-programmable gate arrays (FPGAs). A more complete description of the proton CT scanner hardware can be found in Ref. 5.

The MSS measures the residual energy and range of each proton. It is composed of 5 stages of scintillating plastic (UPS-923A, polystyrene), each 36.0×10.0×5.1 cm³. Stages through which protons pass entirely contribute directly to total range, while the stage in which the proton stops measures residual energy, which is converted via calibration into residual range. Integrated light guides interface through photomultiplier tubes (PMTs). PMT signals are digitized on a custom board by fast pipelined analog–digital converters and interfaced to the data acquisition by FPGAs. A detailed description of the MSS can be found in Ref. 6.

2.B. Custom edge phantom

A custom edge phantom (Fig. 1) was designed by the first author and fabricated by Computerized Imaging Reference Systems, Inc. (CIRS), Norfolk, VA. It was designed for measuring the MTF of the proton CT scanner. The phantom has a diameter of 200 mm, a height of 60 mm, and is composed primarily of water-equivalent plastic (CIRS Plastic Water-LR, RSP = 1.007). It contains four groups of rectangular inserts composed of three different tissue-equivalent polymers representing enamel (RSP = 1.770), adult cortical bone (RSP = 1.685), and lung (RSP = 0.217), as well as air (RSP = 0.007). The inserts are completely contained within the body of the phantom and have dimensions of 15×15×45 mm³. The centers of the inserts are positioned at radii of 25, 55, and 80 mm, respectively, such that their innermost and outermost edges are orthogonal to the radius of the cylinder. Three drill holes of 1 mm radius are located 95 mm from the center of the phantom defining the coordinate axes. The first enamel insert is rotated 5° with respect to the x-axis. The angle between adjacent inserts is 30°.

2.C. Edge phantom scans

The proton CT scanner was installed on the fixed horizontal beamline at the Northwestern Medicine Chicago Proton Center (NMPC), and the edge phantom was placed on the rotation stage with the drill holes aligned with the alignment lasers, as shown in Fig. 2. In the first scan, the phantom was rotated in 4° intervals and about 4×10⁶ proton events were acquired at each rotation angle. In the second scan, the phantom was rotated continuously at a rate of one rotation per minute over a period of 7 min, acquiring about 320×10⁶ proton events. The continuous scan corresponds to an internal dose of approximately 1.4 mGy, which was measured using the Catphan CTP554 16 cm acrylic dose phantom (The Phantom Laboratory, Greenwich, NY), using equivalent scanning parameters. Scaling this proportionally gives a dose estimate of approximately 1.6 mGy for the stepped scan.

2.D. Monte Carlo simulations

The edge phantom was also simulated using TOPAS 2.0, which is based on GEANT4 version 10.01.p02, using the standard physics activation for TOPAS. The simulation was performed using two different levels of idealization of proton CT scanning: In the first simulation, a 200 MeV uniform parallel beam was incident on an idealized proton CT system where the silicon tracking detectors were replaced with sensitive areas composed of air, thus eliminating MCS in the tracking detectors, and the entrance and exit coordinates were determined exactly. The multistage scintillator was eliminated and instead the energy of the protons was evaluated at the front and the rear trackers, respectively; the energy loss was converted to WEPL via a calibration procedure that correlated energy loss with the water equivalent thickness (WET) of a series of degraders with varying WET placed in the simulated proton CT scanner. In the second simulation, silicon trackers equivalent to those used in the prototype scanner were restored; however, exact measurement of tracking coordinates, the idealized energy detector, and the uniform parallel beam remained the same. The measurement uncertainty due to the strip pitch of the SSD was then simulated by adding random Gaussian noise with a width of 228 μm/√12 to the tracking coordinates in the output from the simulation with the silicon tracker. The simulated scans were performed in 4° steps and
the resulting data sets contained approximately $140 \times 10^6$ histories each.

2.E. Image reconstruction

2.E.1. Description of the algorithm

The software for the reconstruction of images was executed on a workstation equipped with two dual-core central processing units, 8 GB of RAM, and an EVGA GeForce GTX680 GPU. The data input to the reconstruction software contained proton tracker coordinates and the WEPL for each proton.

Image reconstruction was accomplished first by projecting the entry and exit vectors measured by the silicon trackers onto the reconstruction volume, and then binning the WEPLs of these protons into a sinogram according to the midpoint of the straight-line path between the points at which the protons enter and exit the reconstruction volume. The distribution in each bin was then analyzed and data cuts were performed in order to eliminate protons that fell outside of $3\sigma$ from the central WEPL value of the distribution. These cuts are effective in minimizing errors from protons that undergo hadronic interactions in the scanner or phantom.

The resulting sinogram was then passed through a Shepp–Logan filter and was used as input to the Feldkamp–Davis–Kress (FDK) algorithm for a 3D filtered back projection (FBP). The FBP image was used both for boundary detection and as a starting point for the subsequent iterative reconstruction. For the iterative reconstruction, the diagonally relaxed orthogonal projection (DROP) onto convex sets was used. This DROP algorithm was further enhanced by interleaved superiorization of the total variation (TV) of the reconstructed image. Details of this DROP-TV superiorization (TVS) algorithm have been described elsewhere.

2.E.2. Implementation of the algorithm

The simulated data sets contained fewer than half as many proton histories as the experimental data sets so in order to keep the number of reconstructed histories constant, the data from the stepped scan and continuous scan were reduced such that the total number of histories that entered reconstruction, after applying $3\sigma$ cuts, was approximately the same for the both the simulated and experimental scans. This was done by randomly selecting a fixed percentage of the events from each projection angle in the experimental scans. For the continuous scan, the incoming angles of the protons were calculated using the time stamps of the events and the known angular speed of the rotation stage; the angles were then binned into $1^\circ$ bins for the FBP.

Twenty iterations of DROP-TVS were performed using two hundred blocks and a relaxation parameter of 0.20. These parameters were chosen based on the results of a comprehensive comparison of many parameter combinations. Images were reconstructed using a pixel size of 0.625 mm and a slice thickness of 2.5 mm.

2.F. Determination of MTFs

The MTF of proton CT has several individual components: the frequency response of the tracking detector measurement, the MCS in the object, and the reconstruction processes. The total MTF of the system is the product of these components. Thus, the MTF of the proton CT scanner can be modeled in the following way:

$$\text{MTF}_{\text{total}} = \text{MTF}_{\text{MCS}} \times \text{MTF}_{\text{detector}} \times \text{MTF}_{\text{recon}}. \quad (1)$$

The MTF was determined from the ESF using the oversampling method of Mori and Machida,\textsuperscript{13,21,22} for characterizing the spatial resolution in CT. The high-contrast materials were juxtaposed to produce a sharp edge with a slight angle, $\alpha$, with respect to the principal axes ($x$, $y$). In general $\tan \alpha$ should not be equal to an integer, or to the ratio of two small integers, in order to ensure that all possible regions of the pixel can be sampled, as the oversampling method requires. The present phantom was constructed using $\alpha = 5^\circ$, $35^\circ$, $65^\circ$, etc. The determination of the MTFs was accomplished using Python version 2.7 with the NumPy and SciPy modules imported.\textsuperscript{23}

Sampling was performed orthogonal to the edge along a central 10 mm segment of the insert edge at 0.625 mm intervals. The resulting ESFs for a particular insert were
overlaid by aligning the 50% values of each ESF in order to yield an oversampled ESF. For noise suppression, the oversampled ESF was rebinned using a bin size equal to the pixel size and a bin pitch equal to one fourth of the pixel size. This was performed for each insert at six different orientations of the phantom. The oversampled ESFs for the six phantom orientations were again overlaid by aligning the 50% values and were rebinned again. A LSF was obtained by linearly interpolating the twice-rebinned ESF, and then numerically differentiating. The discrete MTF for each slice was obtained by Fourier-transforming the LSF using a discrete Fourier transform. The appropriate corrections, as described by Mori and Machida, were applied to the LSF. A cubic spline was used to define the continuous MTF. This process was performed for each of 14 slices. The discrete MTFs for each slice were then averaged, and a spline was fitted to the data to obtain the average MTF curves. We took the MTF\textsubscript{10\%} to be the spatial resolution, which is the spatial frequency at which the MTF has fallen to a value of 10\%.

3. ANALYTIC ESTIMATE

The 2008 paper by Schulte et al.\textsuperscript{10} described the formalism for determining the MLP and its transverse position uncertainty for protons traversing a water slab of constant thickness. The uncertainty curves for 200 MeV protons traversing $W = 5$, 10, 15, 20, and 25 cm of water, which were calculated using this model, are shown in Fig. 3, which plots the uncertainty $\sigma_W(u)$ versus the depth $u$ in the object.

In order to obtain an analytic estimate of the spatial resolution due to MCS in the phantom, we considered a simplified model of a homogeneous, cylindrical water phantom in a parallel beam. The MLP uncertainty at a depth, $u$, along the proton’s trajectory was obtained from the function $\sigma_w(u)$. We made the simplifying assumption that the spatial resolution is primarily limited by the proton tracks that pass parallel to the edge, since the uncertainty of these tracks is transverse to the edge. Given this assumption, there are two primary factors that contribute to the spatial resolution at a particular edge: the length, $W$, of the proton path, and the positions along the path $u$ at which the uncertainty in the MLP is evaluated.

Figure 4 shows that the proton tracks that contribute to the azimuthal resolution traverse the same thickness of water, $W = 2\sqrt{R^2-(w/2)^2}$, where $w$ accounts for the width of the inserts in the physical phantom, regardless of the radial position of the edge. Protons contributing to the radial resolution traverse varying thicknesses of water depending on the position of evaluation such that $W = 2\sqrt{R^2-r^2}$, where $r$ is the radial displacement of the edge from the center of the phantom. It is apparent that the proton tracks that contribute to azimuthal resolution are typically longer than those which contribute to radial resolution. However, the position along the path at which the MLP uncertainty is evaluated is closer to the periphery of the phantom where the MLP is less uncertain, whereas the tracks contributing to radial resolution are evaluated at $W/2$, near the maximum of the MLP uncertainty curve for any $W$.

In addition, for the azimuthal measurement, the MLP uncertainty curve is evaluated at the two points,


\[ u_\pm = \sqrt{R^2 - (w/2)^2} \pm \sqrt{r^2 - (w/2)^2}, \]

since the cylindrical symmetry permits protons to traverse the phantom in either direction. Although the sum of two Gaussian distributions is not strictly Gaussian, it was found by using a random Gaussian number generator that, when the values of uncertainty relevant to this study, the resulting distributions were close to Gaussian, with a width approximately equal to the arithmetic mean of the two uncertainties, and so this was used to approximate the average uncertainty in the azimuthal direction.

The evaluated uncertainty, \( \sigma_0 \), was taken to be the width of the PSF at the position \( r \). Since the MTF is the Fourier transform of the PSF, the width of the MTF is related to \( \sigma_0 \) by

\[ \sigma_{MTF}^2 = \frac{1}{4\pi^2\sigma_0^2}. \] (2)

Therefore, an estimate of the MTF\(_{10\%}\) due to the MCS in the phantom may be obtained by evaluating the 10\% point of the Gaussian function \( f(k) = e^{-2\pi^2\sigma_k^2} \).

4. RESULTS AND DISCUSSION

Figures 5(a) and 5(b) show slices of the reconstructed edge phantom from the fully idealized simulation (in which silicon trackers were replaced with air) and the stepped experimental scan, respectively. A slice thickness of 2.5 mm and a pixel size of 0.625 mm were used, and the image size is 354 × 354 pixels.

The MTF\(_{10\%}\) of the ideal simulation was compared with our analytic estimate. These results are presented in Fig. 6. The left hand panel plots the radial MTF\(_{10\%}\) as a function of the path length \( W \), whereas the right hand panel plots the azimuthal MTF\(_{10\%}\) as a function of the shorter of the two displacements of the position of evaluation from the MLP endpoints, \( u_- \). The solid curves represent the analytic estimate of spatial resolution imposed by MCS. The orange diamonds and blue circles represent the radial and azimuthal spatial resolution, respectively, of the image reconstructed using 0.625 mm pixels [Fig. 5(a)]. The data points were obtained by taking the average of all the edges at the same radial position in the phantom. The error bars represent the range of the values. The dashed curves on the azimuthal plot represent the analytic estimates associated with proton whose MLP uncertainties are evaluated at \( u_- \) and \( u_+ \), respectively. Those evaluated at \( u_- \) in general, have smaller MLP uncertainty than those evaluated at \( u_+ \) due to increased MCS at lower energies.

The radial spatial resolution of the image reconstructed from the fully idealized Monte Carlo simulation ranges from 6.31 ± 0.36 lp/cm for \( W = 197 \) mm to 13.79 ± 0.036 lp/cm for \( W = 97 \) mm. The azimuthal spatial resolution ranges from 6.99 ± 0.23 lp/cm at \( u_- = 75 \) mm to 11.20 ± 0.26 lp/cm at \( u_- = 20 \) mm. The simulation agrees well with the model near the center of the phantom, when the radial position of the edge is less than about 60 mm. This suggests that the simplified model (Sec. 3) describes the effect of MCS on spatial resolution well. For more peripheral inserts, the estimate for the spatial resolution imposed by MCS diverges rapidly from the values measured in simulation, which indicates that MCS does not limit radial spatial resolution for \( W < 130 \) mm or azimuthal spatial resolution for \( u_- < 40 \) mm. For smaller values of \( W \) and \( u_- \), near the periphery of the phantom, the limiting behavior is likely due to binning of the FBP, as well as the size of the pixels in the image.

Additional degradation of spatial resolution may be introduced by inhomogeneities in the phantom, and by the reconstruction process. The MLP model used in image reconstruction assumes a uniform material (water), which is a reasonable approximation for soft tissues in the absence of large inhomogeneities, such as bone and air. When high contrast inserts are present along the proton path, the MLP of the proton is incorrectly approximated and errors are propagated along the proton’s path through the phantom, causing an overall reduction in spatial resolution.

There may still be room for improvement in spatial resolution, which may be achieved by improving the MLP model. This should include knowledge of the positions of high and low density materials after the initial FBP reconstruction has been performed. Improving the quality of the initial iterate for DROP-TVS may also lead to higher spatial resolution. In
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Fig. 6. Plots comparing the analytic estimates (solid and dashed curves) for radial and azimuthal spatial resolution with the results of a fully idealized simulation (data points). The radial spatial resolution (left) is plotted as a function of the total (straight line) path length through the phantom ($W$), whereas the azimuthal spatial resolution (right) is plotted as a function of the shorter of the two displacements, $u_-$, of the position of evaluation from the MLP endpoints. The dashed curves on the azimuthal plot represent the analytic estimates associated with protons whose MLP uncertainties are evaluated at $u_-$ and $u_+$, respectively. The pixel size of the images is 0.625 mm. The average MTF$_{10\%}$ of enamel, cortical bone, lung, and air are plotted as points, with the error bars indicating the range of measurements.

In addition, there are several new algorithms in development that can be applied to proton CT, which should also be thoroughly evaluated and compared with our DROP-TVS algorithm.$^{8,9,24,25}$

Plots of the MTF$_{10\%}$’s for all edges for the stepped experimental scan [Fig. 5(b)] are given in Fig. 7. The plots show that the spatial resolution in the radial direction ranges from $5.11 \pm 0.61$ lp/cm for $W = 197$ mm to $8.58 \pm 0.50$ lp/cm for $W = 97$ mm. In the azimuthal direction, the spatial resolution ranges from $5.37 \pm 0.40$ lp/cm at $u_- = 75$ mm to $7.27 \pm 0.39$ lp/cm at $u_- = 20$ mm. The error bars in this plot represent the statistical uncertainty (1 standard deviation) on the average value of the MTF$_{10\%}$ and were obtained by averaging the MTF$_{10\%}$'s evaluated for each z-slice of the phantom. In addition, Fig. 8 shows the radial and azimuthal MTFs for the enamel inserts. The dashed gray lines indicate the MTF$_{10\%}$. The MTFs of the three other materials look similar.

Fig. 7. Plots of the MTF$_{10\%}$’s for the radial MTFs for the six radial edges for each of the four material inserts (left), and for the azimuthal edges at three radial positions for each of the four material inserts (right) for the image reconstructed from experimental data [Fig. 5(b)] using 0.625 mm pixels and 2.5 mm slices. The radial MTF$_{10\%}$’s are plotted versus the path length, $W$, whereas the azimuthal MTF$_{10\%}$’s are plotted versus the shortest displacement from the MLP endpoints, $u_-$.
As expected, MTF\textsubscript{10\%}’s show strong dependence on path length and position of evaluation, for radial and azimuthal edges, respectively, and little material dependence. Although the effect is not large, in the radial plot, it appears that the high density inserts, enamel and cortical bone, tend to have consistently lower spatial resolution than the low density inserts, air and lung. This may be understood in terms of the amount of scattering the different density materials cause, which affects the MLPs in different ways. Scattering will be largest for the high-density materials (bone and enamel), and for lower proton energies at smaller radial displacements. The material dependence of the azimuthal edge resolution is even less pronounced than that of the radial edge distributions. This may be because the geometry of the phantom is such that high and low density inserts are always opposite to one another, and so reduced scattering in the low density inserts may be counteracted by increased scattering in the high density inserts.

Figure 9 shows a comparison between the MTF\textsubscript{10\%} values of the most ideal simulation, the simulation with the realistic silicon tracker, the stepped experimental scan and the continuous experimental scan. The plot shows strong agreement between the idealized simulation with the silicon tracker and the two experimental scans. Evidently, replacing the ideal tracking detectors with silicon detectors significantly reduced the spatial resolution overall, ranging from a reduction of 20% for $W = 197$ mm to 37% for $W = 97$ mm. This is because the MLP model assumes exact knowledge of the entry and exit points from the phantom, whereas due to MCS and point resolution uncertainty in the silicon trackers, the MLP...
endpoints are fairly uncertain. This increases the uncertainty at every point along the MLP curve, with the most significant increase occurring near the endpoints, where the uncertainty is zero in the ideal case. This results in the largest reduction in spatial resolution occurring near the periphery of the phantom, (small W and $u_\perp$) as is indicated by the plots.

In addition, the azimuthal resolution is reduced by about 5% more than the radial resolution near the periphery of the phantom. This is expected because the protons that contribute the largest uncertainty at these points have much larger W than their radial counterparts, resulting in the azimuthal protons having less residual energy when they enter the rear tracker. Because of this, they are subject to a greater amount of MCS in the rear tracker than protons with higher residual energy.

As indicated by Fig. 9, the range of MTF values from continuous experimental scan, and simulated and experimental stepped scans overlapped for both radial and azimuthal resolutions. This means that a stepped scan with 4° steps gives equivalent results in terms of spatial resolution, which is important when the proton beam is pulsed (as is the case for synchrotrons) and thus continuous scans are not optimal.

5. CONCLUSIONS

We have characterized the spatial resolution of a prototype proton CT system by comparing an analytic estimate with two simulated systems at various levels of idealization, and with two experimental scans. We have demonstrated an understanding of how each component of the prototype system affects the spatial resolution of a proton CT image. Comparison of the fully idealized simulation with the analytic estimate suggests that near the center of the phantom, the spatial resolution is limited by MCS; however, for path lengths of less than approximately 130 mm, other factors are more limiting than MCS. It may still be possible to further improve spatial resolution by using an improved MLP estimate and/or an improved initial iterate from DROP-TV. Comparing the more realistic simulation and the experimental results with the fully idealized simulation reveals that MCS in the silicon detectors is the largest contributor to the degradation of spatial resolution, which poses a convincing case for a more transparent proton tracking system. Since the silicon wafers were left over from a previous and unrelated project, their design was not optimized for application in proton CT. A preferable design may use 200 µm thick double sided silicon detectors, which would reduce the amount of material by a factor of 4. The strip pitch of the SSDs also results in a small reduction in spatial resolution due to uncertainty in the coordinate measurement. This could be reduced by using SSDs with a finer pitch, but it is a fairly insignificant effect compared with the thickness of the detectors and would increase the complexity of the detector readout.

In order to further improve spatial resolution of proton CT, we suggest to develop more transparent tracking detectors, and to implement improved path estimation algorithms. These may include starting from a more advanced FBP algorithm, and the incorporation of information about heterogeneities in the MLP formalism of the subsequent iterative reconstruction. Finally, the development and availability of higher-energy medical accelerators for protons as well as for heavier ions will lead to images with improved spatial resolution.

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CONFLICT OF INTEREST DISCLOSURE

The authors have no COI to report.

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