A Novel Imaging Mode Based on X-ray CT Prior Image and Sparsely Sampled Projections for Rapid Clinical Proton CT

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Abstract

Proton computed tomography (CT) has a distinct practical significance in clinical applications. It eliminates 3–5% errors caused by the transformation of Hounsfield unit (HU) to relative stopping power (RSP) values when using X-ray CT for positioning and treatment planning systems (TPSs). Following the development of FLASH proton therapy, there are increased requirements for accurate and rapid positioning in TPSs. Thus, a new rapid proton CT imaging mode is proposed based on sparsely sampled projections. The proton beam was boosted to 350 MeV by a compact proton linear accelerator (linac). In this study, the comparisons of the proton scattering with the energy of 350 MeV and 230 MeV are conducted based on GEANT4 simulations. As the sparsely sampled information associated with beam acquisitions at 12 angles is not enough for reconstruction, X-ray CT is used as a prior image. The RSP map generated by converting the X-ray CT was constructed based on Monte Carlo simulations. Considering the estimation of the most likely path (MLP), the prior image-constrained compressed sensing (PICCS) algorithm is used to reconstruct images from two different phantoms using sparse proton projections of 350 MeV parallel proton beam. The results show that it is feasible to realize the proton image reconstruction with the rapid proton CT imaging proposed in this paper. It can produce RSP maps with much higher accuracy for TPSs and fast positioning to achieve ultra-fast imaging for real-time image-guided radiotherapy (IGRT) in clinical proton therapy applications.

Full Text

Preamble

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Abstract

Proton computed tomography (CT) holds significant practical value in clinical applications, as it eliminates the 3–5% errors associated with converting Hounsfield units (HU) to relative stopping power (RSP) values when using X-ray CT for positioning and treatment planning systems (TPSs). With the advent of FLASH proton therapy, there is a growing demand for accurate and rapid positioning in TPSs. This study proposes a novel rapid proton CT imaging modality based on sparsely sampled projections. A proton beam is accelerated to 350 MeV using a compact proton linear accelerator (linac). GEANT4 simulations were conducted to compare proton scattering at 350 MeV versus 230 MeV. Since sparsely sampled data from 12 projection angles is insufficient for reconstruction alone, X-ray CT is employed as a prior image. The RSP map generated from X-ray CT conversion was constructed through Monte Carlo simulations. Considering most likely path (MLP) estimation, the prior image-constrained compressed sensing (PICCS) algorithm was used to reconstruct images from two different phantoms using sparse 350 MeV parallel proton beam projections. The results demonstrate the feasibility of proton image reconstruction using the proposed rapid proton CT imaging approach. This method can generate RSP maps with significantly improved accuracy for TPSs and enable fast positioning, achieving ultra-rapid imaging for real-time image-guided radiotherapy (IGRT) in clinical proton therapy applications.

Keywords: Proton CT, real-time image guidance, image reconstruction, proton therapy

I. Introduction

Proton therapy currently accounts for approximately 85% of hadron therapy cases in daily tumor treatments conducted at high speed [?]. Owing to its Bragg peak characteristics, most energy is released at the lesion site, thereby minimizing damage to healthy tissues and cells. However, both proton and photon therapies currently rely on X-ray CT for treatment planning systems, where HU values are converted to RSP maps. This conversion process introduces intrinsic errors of 3–5% [?], which can have serious consequences for sensitive organs and blood vessels. Proton CT can address this error and reduce uncertainties associated with organ movement when patients are transferred to the treatment room.

FLASH therapy has emerged as a new focus in tumor radiotherapy due to its ultrahigh dose rate properties (FLASH > 35 Gy/s) [?], imposing more stringent requirements for precise tumor localization and improved image-guided radiotherapy (IGRT). The first domestic proton therapy device has entered clinical use at the Ruijin Hospital Proton Therapy Center, Shanghai Jiaotong University School of Medicine. The proton CT imaging project is planned for the fifth treatment room for experimental validation, creating a pressing need for clinically relevant rapid proton CT imaging.

Although research teams worldwide are developing proton CT, it remains in the experimental stage and has not yet been adopted clinically due to various limitations. For example, experimental setups typically rotate phantoms to obtain projection information from different angles—a scheme not applicable to patients in clinical settings [?]. While clinical proton therapy uses omnidirectional irradiation at 180 angles, this approach is impractical for proton CT due to the lengthy rotation time required. Consequently, achieving rapid proton imaging to obtain accurate RSP maps for clinical TPSs has become a focal point of research. Moreover, conventional gantries employ heavy magnets; integrating proton CT into a rotating gantry would result in prohibitively long irradiation times due to rotational inertia. To achieve accurate RSP reconstruction in a very short time, combining proton CT with X-ray CT presents a viable solution. Reducing the number of irradiation angles represents the most direct and efficient approach to achieving rapid proton imaging.

Meanwhile, a superconducting gantry design distinct from conventional therapy gantries has been proposed [?] that can increase proton irradiation speed. The energy range of existing proton therapy (70–235 MeV) is insufficient for proton imaging [?]. Compared with 230 MeV protons, 350 MeV protons exhibit less scattering and deposit smaller doses in the human body. This study presents a novel proton CT imaging modality based on sparsely sampled projections using X-ray CT as a prior image for rapid clinical proton CT. The proton beam is accelerated to 350 MeV by a compact proton linac. This high-energy proton beam keeps the Bragg peak completely outside the human body and offers major dose advantages compared with X-ray CT. The feasibility of increasing proton energy from 230 MeV to 350 MeV has been verified [?], and a similar acceleration structure has been cold-tested, confirming the design's viability [?, ?].

To achieve sparse-angle proton imaging, X-ray CT images serve as prior images. The X-ray CT is converted to an RSP map via an HU-RSP transformation curve for initial iterations, followed by image reconstruction using 12 parallel proton beam irradiations. The X-ray CT can be any previous scan the patient underwent before proton therapy, provided it is consistent with the proton CT to supply sufficiently accurate prior information. The choice of 12 imaging angles is motivated by a proposed static superconducting therapy gantry based on 12 superconducting coils [?], where coil positions correspond to proton beam irradiation directions. Proton therapy can be rapidly realized through ultra-fast beam allocation and ultra-fast proton scanning technology using an advanced deflector with variable polarization. Fast scanning systems require approximately 10 s to scan an energy layer [?], and acquisition at each angle can be completed within 1 ms. Thus, 12-angle proton projections would finish within 12 ms. Rapid proton CT image reconstruction with an X-ray CT prior can be achieved through hardware combined with efficient proton detection technology, with the entire proton CT process planned to complete within several hundred milliseconds. After imaging, the gantry can quickly adjust energy to perform proton therapy. The static superconducting gantry is designed to perform proton FLASH therapy within 100 ms [?], making the total time for proton CT and subsequent FLASH therapy approximately 1 s [?]. This approach not only improves proton CT speed but also reduces TPS errors caused by organ movement, representing significant clinical progress toward realizing IGRT in proton therapy.

II. Methods

The proton beam energy can be enhanced to 350 MeV using an S-band high-gradient acceleration structure. Information about protons passing through the object is tracked and recorded for reconstruction algorithms. This section introduces the principles and reconstruction algorithms for this imaging modality.

2.1 Principle of the 350 MeV Proton CT Imaging Mode Based on X-ray CT as Prior Image

In this study, the proton beam energy was increased from 230 MeV to 350 MeV. Since sparse scanning at 12 angular projections is insufficient for accurate proton image reconstruction, an X-ray CT image is used as a prior image. To utilize X-ray CT for reconstruction, the transformation relationship between HU and RSP is typically determined experimentally using clinical devices. Different phantoms of known materials are irradiated with X-rays and protons, and the data are interpolated and fitted according to HU values and RSP to obtain the transformation relationship [?, ?]. The resulting HU-to-RSP map enables rapid proton image reconstruction to obtain more accurate RSP maps for TPSs. This paper uses GEANT4 simulation for verification, with X-ray CT and proton CT simulated separately.

The specific process is shown in Fig. 1 [Figure 1: see original paper]. The proton CT system consists of two main parts, as illustrated in Fig. 2(a) [Figure 2: see original paper]. The proton beam enters from the left side. Tracking detectors record the transverse position upstream and downstream of the patient, while a residual range detector determines proton energy loss, enabling calculation of the water-equivalent path length (WEPL) through calibration [?]. The WEPL value is defined as the integral of the material's RSP along the proton path through the object. RSP is defined as the ratio of the stopping power (SP) of the specific material to that of water [?]. Assuming the energy of an incoming proton equals that of a proton ejected from the accelerator, this information serves as the basis for RSP map reconstruction [?].

The proton beam energy can be increased from 230 MeV to 350 MeV using S-band high-gradient accelerating technology. The S-band high-gradient acceleration structure meets requirements for compact facilities and can be used in single-treatment rooms and for FLASH proton therapy [?]. The accelerated proton beam can be transmitted directly into the treatment room for imaging and therapy. Simulations have demonstrated sufficient capture rate [?]. Increasing beam energy using the linac with cold-state technology shortens the linear acceleration section but requires more complex control technology [?, ?, ?]. The superiority of the 350 MeV proton beam was verified by passing different proton beams through various phantoms using 230 MeV and 350 MeV proton pencil beams.

The simulation phantoms included: (a) a 15 cm water cube and (b) a 15 cm radius sphere comprising five different materials. Sigma values (standard deviation from Gaussian fit of the proton beam) were recorded every 10 mm. As shown in Fig. 2(b) [Figure 2: see original paper], the 350 MeV proton beam exhibited smaller sigma values than the 230 MeV beam in both phantoms, indicating less extensive scattering. When 350 MeV protons pass through multi-material phantoms, they scatter more than when passing through a water cube. For sensitive human organs, high-energy proton CT provides better protection. Additionally, the equivalent water depth of the 350 MeV proton beam is approximately 653 mm, ensuring the Bragg peak falls outside the human body to reduce dose. The proton CT dose was approximately 30 mGy in simulations, lower than that of X-ray CT, underscoring the importance of 350 MeV proton beams for imaging.

2.2 Proton Image Reconstruction Algorithms

Similar to X-ray CT, proton image reconstruction algorithms fall into two categories: filtered back projection (FBP) and iterative methods. However, FBP requires sufficient sampling angles. To achieve fast proton CT imaging, iterative reconstruction was selected for high-quality images. Three iterative methods were considered: algebraic reconstruction technique (ART), compressed sensing (CS), and prior image-constrained compressed sensing (PICCS). The basic principle involves selecting an initial matrix for iteration until final criteria are met, which can be either a standard value or a set number of iterations. The methods differ in their iterative equations.

ART is one of the simplest methods for reconstructing images from measured projections. The RSP value is represented by x, and the consistency condition should be satisfied [?, ?]:

(2.2)

Iterative reconstruction is consistent in the least-squares sense. In proton CT applications, vector b describes the measured WEPL values, and A represents the system matrix—an M $\times$ N matrix whose elements indicate the contribution of the ith projection to the jth pixel. Typically, aij represents either the intersection length between the ith proton path and jth pixel or the intersection probability between each mesh j and path i. The calculation of projection matrix A directly affects overall algorithm reconstruction speed.

Sparse images are reconstructed iteratively by minimizing the 1 norm, defined as the summation of absolute function values, while satisfying equation (2.2). In PICCS, the sparseness of the target image is defined relative to a prior image through subtraction operations, where FP represents the prior image. Sparsifying transforms, including discrete gradient transform (F - FP) and wavelet transforms, are applied to the subtracted image. This study used the total variation term TV(F), defined as the 1 norm of the discrete gradient image. The parameter α ($0 \leq \alpha \leq 1$) is selected based on similarity between the prior image FP and reconstructed image F. When $\alpha = 0$, PICCS reduces to the CS algorithm [?, ?].

To compare reconstruction results, the reconstructed image was quantitatively compared against the actual phantom image using normalized mean square distance d and normalized mean absolute distance r, calculated for each algorithm [?, ?]. RSP accuracy was expressed as Diff:

(3.1)
(3.2)
(3.3)

where tu,v and ru,v represent pixel RSP values at row u and column v in the original and reconstructed phantom images, respectively, and tmean is the average pixel value of the object. The image contains N $\times$ N pixels. Smaller d values indicate better image quality. RSPreal represents the true RSP value, and RSPrec represents the reconstructed RSP value.

III. Results

This study constructed proton CT and X-ray CT platforms based on Monte Carlo simulations. The proton CT simulation software platform comprised GEANT4 simulations [?], WEPL calibration, and image reconstruction. GEANT4 simulations required determining specific three-dimensional models and parameters, including detector parameters, proton beam properties, phantom characteristics, and physical processes. A critical step in proton CT involves determining energy loss when protons travel fixed distances within objects [?]. After data fitting, two-dimensional proton projection data can be obtained using quadratic relationships. Similarly, detector parameters and X-ray energy must be set to reconstruct X-ray images based on attenuation through objects. Proton projection data (acquired at 12 angles) and the HU-RSP map were used for image reconstruction.

3.1 Most Likely Path Estimation for Accurate Proton Positioning

Proton image reconstruction requires calculating the likely proton path within the phantom based on appropriate multiple Coulomb scattering theory, given entry and exit positions. The position and angle detector system accurately measures each proton's position and direction, enabling path estimation through the object.

Proton path estimation comprises three types: straight line path (SLP), cubic spline path (CSP), and most likely path (MLP). SLP is the simplest estimation, defined by the line between entry and exit positions. CSP fits smooth curves to the two endpoints using known positions and tangential directions at entrance and exit, offering mathematical simplicity over MLP. However, MLP provides the best statistical estimate of the path curve for protons traversing homogeneous media. Estimated paths are incorporated into iterative reconstruction algorithms.

Internal paths follow probability distributions, and MLP can be defined as a maximum likelihood problem in Bayesian statistics [?, ?]. With incoming proton position and direction set to A (0, 0, 0) and exit position and direction set to B (u1, t1, 1), the MLP ymlp in a uniform object is given by [?]:

(3.4)
ymlp = (R0y0 + R1

Every term in the equation has a detailed calculation method [?]. To calculate scattering matrix elements, proton speed relative to light velocity and proton momentum at different depths, $1/\beta^2 p^2$, were derived using GEANT4 simulations. In these simulations, 350 MeV protons traversed a uniform 200 mm water cube, approximated by a fifth-degree polynomial with fitted coefficients shown in Table 1 . As shown in Fig. 3 [Figure 3: see original paper], information from two protons was randomly recorded from a simulation, and CSP, MLP, and associated error envelopes were calculated (protons entered from the right). The figure shows MLP and CSP are closer to the real path than SLP. Root-mean-square error (RMSE) calculations in the central region yielded values of 0.764 mm, 0.443 mm, and 0.422 mm for SLP, CSP, and MLP, respectively. The SLP RMSE was approximately twice that of MLP, while MLP RMSE was slightly smaller than CSP, confirming MLP provides more accurate proton path estimation.

3.2 Algorithm Optimization to Improve Reconstruction Performance

Through continuous image reconstruction improvement, the imaging mode using the HU-RSP map as a prior image was preliminarily verified using the aforementioned algorithms. A spherical water phantom (25 cm diameter) was constructed containing two smaller spheres of different materials and sizes. Projection data were collected in GEANT4 using a 350 MeV parallel proton beam at 12 angles. For verification, standard image values were uniformly increased to serve as a prior image representing the RSP-converted X-ray CT. Figures 4(a), 4(b), and 4(c) show reconstruction results using ART, CS, and PICCS, respectively. Visually, the ART-reconstructed image in Fig. 4(a) [Figure 4: see original paper] exhibits obvious artifacts. In contrast, PICCS reconstruction in Fig. 4(c) shows considerably improved image quality without artifacts in the largest circle.

The d and r values for all three algorithms were calculated and plotted as functions of iteration number in Fig. 4(d) [Figure 4: see original paper]. Longitudinally, values decrease with increasing iterations, indicating improving reconstruction quality until reaching a steady state. Horizontally, PICCS yields lower d and r values, demonstrating significantly improved image quality. After 100 iterations, ART, CS, and PICCS achieved d values of 0.174, 0.147, and 0.131, respectively, and r values of 0.1, 0.098, and 0.086, respectively. When rotated 45° clockwise, line 262 RSP values were compared as shown in Fig. 4(e) [Figure 4: see original paper]. The yellow solid line represents true RSP values, and the yellow dashed line represents prior image values. The ART result shows more noise, while the CS result performs poorly at material edges. The PICCS result agrees well with initial values.

In the PICCS algorithm, α selection is critical [**?**, **?**] and depends on prior image quality. Since RSP deviation is uniformly distributed in this prior image and accurate boundary information is provided, $\alpha = 0.95$ was chosen after parameter tuning. For poor-quality prior images, α would take smaller values. The RSP distribution image obtained through HU-RSP transformation currently provides relatively accurate prior information, necessitating α values that cannot be too small. However, the specific α value must be determined through tuning under different experimental conditions. Based on d and r calculations, PICCS yields superior results when using the proposed method with a prior image and 12-angle proton irradiation. Monte Carlo simulations will be used to obtain X-ray and proton CT images of different phantoms to verify the proposed imaging mode.

Many factors affect proton CT image reconstruction and final RSP accuracy. Semiconductor detectors can achieve micron-level resolution, which is sufficient for proton CT. This paper analyzes the energy resolution of residual energy detectors and angle accuracy of the integrated proton CT gantry. In GEANT4, a 350 MeV parallel proton beam collected 12-angle projection information from Phantom2, and the PICCS algorithm was used for reconstruction.

First, the allowable error range of the residual energy detector was analyzed. The deviation between proton energy used for reconstruction and real residual energy was set to 0, \$±\$0.5, \$±\$1, and \$±\$2 MeV. The d and r values of reconstructed images under different conditions were calculated, with results shown in Table 2-1 -1. Comparing d and r results shows that reconstructed RSP accuracy is close to true values for \$±\$0.5 and \$±\$1 MeV cases, indicating that energy resolution in the 0.3–0.6% range enables error tolerance in image reconstruction.

Second, the allowed error range of gantry angle was verified through GEANT4 simulation. Deviations between reconstruction algorithm projection angles and actual simulation irradiation angles were set to 0°, +0.2°, +0.5°, and +1° to simulate therapy gantry angle accuracy. The 0° deviation represents the standard case, with quantitative results shown in Table 2-2 -2. Comparison reveals that d and r values are larger than the standard case for 0.5° and 1° deviations. Therefore, reconstruction results remain within allowable error range when gantry angle deviation is within 0.2°. The therapy gantry angle accuracy of the first domestic proton therapy demonstration device is within 0.2°, providing favorable experimental conditions for subsequent proton CT experiments.

3.3 Rapid Proton CT Imaging Based on X-ray CT as Prior Image

Monte Carlo simulations were used to generate X-ray CT images. In these simulations, projection data with relative errors $\leq 0.75\%$ were collected at 180 angles over half a circle using a 60 keV parallel photon beam and reconstructed using traditional FBP. The detector layer comprised 400 pixels of 1 mm width. PICCS reconstruction yielded better results under the same conditions.

Two phantoms were selected for simulations. Figure 5(a) [Figure 5: see original paper] shows a phantom composed of four different materials: the largest sphere is water (radius = 12.5 cm) containing three smaller spheres of different materials (radii = 2, 1.5, 1, and 0.5 cm) made of Teflon, PMMA, and air. Figure 5(b) [Figure 5: see original paper] shows another phantom composed of five different materials with densities closer to human soft tissue: POM, PMP, PMMA, and Epoxy [**?**]. All diameters equal 1.5 cm, and the largest sphere radius is 10 cm. These two phantoms are referred to as Phantom2 and Phantom3.

To verify the feasibility of the proposed rapid proton CT imaging mode, the PICCS method was used for proton image reconstruction with 180, 120, 60, 30, and 12 projection angles, along with FBP using 180 angles. The initial iteration matrix was zero. Corresponding material RSP values and errors were calculated, with results listed in Table 3 . When projection angles are reduced to 30 and 12, Diff values increase significantly, with maximum reconstruction error at 12 angles. While Teflon and PMMA Diff values do not follow a clear trend, this simulation serves as reference and comparison for the proposed imaging mode, demonstrating that the prior-image-based mode with 12 sparse angles can achieve reconstruction results of the same magnitude as those with more angles. Results for 180 and 120 angles are basically within standard value ranges. Due to systematic errors in simulated proton CT, reconstructed RSP values may fluctuate within system error bounds.

Multi-angle proton CT imaging achieves relatively accurate reconstruction but lacks adequate speed due to numerous scanning angles. The HU-RSP images of both phantoms are shown in Figs. 6(a) and 6(c) [Figure 6: see original paper]. For Phantom2, transformed RSP values for Teflon, PMMA, and water were 1.467, 1.081, and 0.972, respectively, with percentage differences from standard RSP values of 18.04%, 6.81%, and 2.8%. Comparisons show larger errors for higher-density materials, similar to experimental conclusions [**?**].

Proton reconstruction was performed using 12 angular projections within 180°. Deviations from true RSP values differed among materials of different densities, indicating that identical algorithmic parameters may not optimize all materials simultaneously. In the PICCS algorithm, excessive increment factors worsen correlation between reconstructed images and projection data. Therefore, different PICCS parameters were set according to material densities. For dense bone media with large conversion errors, fewer TVM settings were used in PICCS. Additionally, because prior image RSP distributions are uneven and noisy, α was set to 0.2 after parameter tuning. Reconstruction results are shown in Figs. 6(b) and 6(d) [Figure 6: see original paper], with Phantom2 and Phantom3 results presented in Tables 3 and 4, respectively. RSPprior represents the RSP from the HU–RSP map, RSPPICCS represents the PICCS-reconstructed value, Diffprior represents the error percentage between HU–RSP map and true values, and DiffPICCS represents the error percentage between reconstructed and true values.

Analysis of Phantom2 reconstruction results shows that with appropriate PICCS parameters, reconstructed RSP values for Teflon, PMMA, and water were 1.769, 1.13, and 1.01, respectively—more accurate than prior image values. Errors between reconstructed and actual values were 1.17%, 2.59%, and 1%. Figure 6(b) [Figure 6: see original paper] shows a clear, high-quality image, verifying the feasibility of the proposed imaging mode and demonstrating that PICCS can yield relatively accurate RSP results. Since Teflon RSP deviation in Phantom2 was larger than other materials, the algorithm was applied to Phantom3 to verify imaging mode feasibility. Phantom3 material densities were all close to human soft tissue.

The constructed RSP values for POM, PMP, PMMA, and Epoxy were 1.340, 0.865, 1.13, and 1.127, respectively, while prior image RSP values were 1.204, 0.844, 1.088, and 1.089. All reconstructed RSP values were closer to actual values, reducing percentage differences from standard RSP values. For instance, POM error decreased from 11.33% to 1.40%, and PMP error decreased from 4.09% to 1.7%. X-ray CT noise impact was smaller compared with HU-to-RSP transformation errors. The two RSP values (RSPprior and RSPPICCS) were fitted as shown in Fig. 7 [Figure 7: see original paper], with correlation coefficient $R^2 = 0.98$. These findings show that HU–RSP transformation yields different deviations for different materials and demonstrate the effectiveness of the fractional RSP algorithm for proton reconstruction using X-ray CT as a prior image. Notably, PMMA reconstruction error using 60-angle proton imaging was also approximately 1.14, potentially attributable to Monte Carlo simulation artifacts.

Comparing the 12-projection imaging mode using the HU–RSP map as the initial iteration with multi-angle proton imaging results reveals that this mode achieves comparable performance. For example, POM RSP error decreased from 11.33% to 1.4%. For Teflon, the proposed imaging mode simulation error was 1.17%, compared with errors of 1.56%, 1.06%, and 1.17% for 180, 120, and 60 angles in PICCS, respectively. Reconstruction from only 12 angles achieves similar results to those from more angles. Data acquisition for 12 angular projections is faster than for sixty or more angles, and the 12-angle reconstruction algorithm completes more quickly, ensuring improved overall proton imaging speed including scanning and reconstruction. Consequently, organ displacement from respiratory motion is also reduced. Simulation experiments verify that the proposed imaging mode can reduce prior image conversion error, with reconstructed RSP error of approximately 2%, confirming the feasibility of rapid proton CT imaging using only 12 angular projections with X-ray CT as a prior image.

The simulation prior image was not processed for artifacts. The PICCS α value of 0.2 was chosen to reduce prior image noise influence, though noise introduction is inevitable. Clinical photon CT is mature and can produce much higher quality images than simulated ones, indicating that the proposed proton CT imaging mode would improve RSP accuracy even with poor-quality prior images, with greater improvements for good-quality priors. This also reduces requirements for error ranges in clinically obtained HU–RSP transformation RSP distribution images. Future algorithm improvements are expected to further reduce reconstructed RSP error. For example, due to multiple Coulomb scattering, protons scatter at object boundaries between different materials, making accurate edge reconstruction vital. Additionally, since the static superconducting therapy gantry can deflect proton beams at different angles through deflection cavities, rapid proton scanning is achievable. Three-dimensional proton CT image reconstruction and further clinical application conditions for the proposed proton CT imaging mode will be studied.

IV. Conclusions

This paper proposed and implemented a novel proton CT imaging mode based on a 350 MeV proton beam using 12 angular projections to achieve rapid proton image reconstruction. This mode combines proton and X-ray CT by using the RSP map converted from acquired X-ray CT as a prior image for TPSs. The MLP was estimated, and the PICCS algorithm yielded superior results for this imaging mode. Monte Carlo simulations of X-ray and proton CT images demonstrated that the imaging mode effectively improves range accuracy compared with HU–RSP transformation maps. A project has been approved to construct a proton CT prototype to verify the principles and imaging outcomes described herein, facilitating a leapfrog development of proton CT from experimental to clinical stages.

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Conflict of Interest Statement

The authors declare no conflicts of interest.

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Note: Figure translations are in progress. See original paper for figures.