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Spatial Resolution Studies for a Prototype Proton CT Scanner

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Abstract—A GEANT4 simulation of the phase II proton CT (pCT) scanner built by the pCT collaboration was used in order to study the spatial resolution of the imaging system. A digital edge phantom composed of water-equivalent polymer with 6 body-equivalent material inserts was created. A modified version of the oversampling method was used in order to construct edge spread functions and modulation transfer functions for materials of varying relative stopping power located at several radial displacements from the center of the phantom. It was determined that for a binning size of $1\text{ mm} \times 2.5\text{ mm}$ and image pixel size of 1 mm^2 , the limiting resolution observed was 50-75% of the Nyquist frequency. This result could be improved to 60-85% of the Nyquist frequency by reducing the binning size for the filtered back projection to $0.5\text{ mm} \times 2.5\text{ mm}$.

Index Terms—proton imaging, tomographic reconstruction of material properties, spatial resolution, modulation transfer function, oversampling method

I. INTRODUCTION

INCREASING use of proton radiation therapy for cancer patients has inspired research into new imaging methods that can improve the accuracy of proton range estimates in radiation therapy planning. X-ray computed tomography (CT) has been shown to give insufficiently accurate maps of relative stopping power (RSP), mostly due to uncertainty in converting CT Hounsfield values to RSP. Since stopping of proton beams near organs at risk (e.g. the spinal cord) requires precise knowledge of proton stopping power, this research has become a high priority for the further development of image-guided proton therapy.

In the phase II pCT scanner, a low-intensity cone beam of 200 MeV protons traverses the patient. Entry and exit positions and vectors and the residual energy are measured for each proton. The residual energy is transformed into water equivalent path length (WEPL) using a calibration method [1]. Based on the WEPL of many protons traversing the object from many directions, one can reconstruct a 3-dimensional

map of RSP. Minimal proton loss and high detection efficiency make pCT a potentially low-dose (m-Gy) imaging modality [2].

Proton CT differs from x-ray CT in several key aspects. In distinction to the absorption of x-rays, where the attenuation factor can vary by a factor of 10 between bone and soft tissue, proton imaging has low contrast, since the RSP differs by only 50% between bone and soft tissue. Furthermore, unlike x-rays which travel in straight lines through the patient or phantom, charged protons are deflected by multiple Coulomb scattering (MCS) events. MCS limits spatial resolution since the proton path deviates from the estimated most-likely path (MLP) by up to several millimeters in anatomical objects encountered in medical proton imaging. Thus, pCT is not intended to produce images of high visual contrast, but rather to create an accurate map of the quantity required for proton radiation therapy treatment planning.

The purpose of this research is to study the spatial resolution of the phase II pCT scanner, and to develop tools that can be used to study the effects of various components of the system on image resolution. At the time of the study the physical phantom was unavailable and so a GEANT4 model of the phase II pCT system was used.

II. MATERIALS AND METHODS

A. Phase II pCT Scanner

The second generation prototype pCT scanner (Fig. 1) consists of two principal components: the silicon tracking system and the energy/range detector. The silicon tracker tracks individual protons. Each of the two trackers consists of 16 silicon tiles arranged into 4 planes, each $400\text{ }\mu\text{m}$ thick, with a total sensitive area of $8.6 \times 34.9\text{ cm}^2$. The upstream and downstream particle telescopes measure the entry and exit vectors of each proton. The tracking planes interface through custom readout ICs [3] and a high-speed data acquisition system based on field-programmable gate arrays (FPGAs) [4]. The multistage scintillator measures the residual energy and range of each proton. It is composed of 5 stages of scintillating plastic (UPS-923A, polystyrene), each $36 \times 10 \times 5.1\text{ cm}^3$. Integrated light guides interface through photomultiplier tubes (PMTs). PMT signals are digitized by fast pipelined analog-digital converters and interfaced to the data acquisition by FPGAs.

B. Reconstruction Software

The software for the reconstruction of images was developed by Scott Penfold, University of Wollongong [5] in col-

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Fig. 1. GEANT4 visualization of the Phase II pCT scanner with a phantom. Single proton is incident from the right and stops in the second scintillator.

laboration with Yair Censor, University of Haifa, and Reinhard Schulte, Loma Linda University, and was modified by Ford Hurley, Loma Linda University. The software was executed on a dual-core workstation equipped with an EVGA GeForce GTX680 GPU. The raw data input into the reconstruction software contained the proton tracker coordinates and the WEPL for each proton.

The radiographic reconstruction made use of the phantom-based coordinate system (x, y, z) and the detector-based coordinate system (t, u, v) which is defined such that u points in the direction of the beam, and v points opposite z . Both systems are right-handed and share a common origin.

The reconstruction software uses a simple algorithm for binning protons according to the straight-line approximation of their path through the phantom. Binning is performed at the proton's projected position at the $u = 0$ plane. The *Feldkamp-Davis-Kress* (FDK) algorithm, the cone-beam version of the filtered back projection (FBP), is used both for boundary detection and as a starting point for iterative reconstruction. For the iterative reconstruction, a method based on diagonally relaxed orthogonal projections (DROP) onto convex sets is used. This method requires knowledge of WEPL and MLP of each proton, forming a large linear system of equations, which is then solved for the unknown RSP object vector. The DROP method has been further enhanced by interleaved superiorization of the total variation in the reconstructed RSP map. Details of the DROP-TVS algorithm have been described elsewhere [6].

C. Phantom Design

Inspired by the study by Seco et al. [7], a phantom (Fig. 2) was simulated for the purpose of determining a modulation transfer function (MTF). The phantom is a water-equivalent polymer cylinder measuring 16 cm in diameter and 6 cm in height. Twelve tissue-equivalent, rectangular inserts, each $15 \times 15 \times 40 \text{ mm}^3$, of varying relative stopping power were inserted in two concentric rings with radii of 35 mm and

Fig. 2. Edge phantom design for spatial resolution studies. The diameter of the phantom is 16 cm and the height is 6 cm. Each insert is $15 \times 15 \times 40 \text{ mm}^3$. The insert materials are as follows: Red = tooth enamel, green = cortical bone, blue = brain, magenta = adult adipose, yellow = lung, cyan = air.

60 mm. The inserts were composed of the materials given in Table I.

TABLE I
MATERIAL INSERTS IN EDGE PHANTOM

Material	Density (g/cm^3)	RSP
Tooth Enamel	2.04	1.796
Cortical Bone	1.85	1.586
Brain	1.07	1.028
Adult Adipose	0.920	0.963
Lung	0.205	0.200
Air	0.0013	0.0011

D. Monte Carlo Simulation

In this study, the GEANT4 version 10.1 [8] toolkit was used to simulate data obtained by the system described above with the edge phantom inserted. Details of the simulation platform were reported by Giacometti et al. [9]. Protons of 200 MeV emerging from the source were scattered by a 1.8 mm-thick lead foil to produce a Gaussian incident beam.

Ninety projections in intervals of 4° were simulated, each containing approximately 610k protons which intersected the reconstruction volume. Of the protons that entered the reconstruction volume, about 60% of them passed statistical data reductions (i.e. 3σ cuts) in angle and in WEPL, leaving approximately 390k protons per projection. A bin size of 1 mm in t and 2.5 mm in v was used for the statistical cuts and the FDK reconstruction. The proton flux through a plane after statistical cuts was therefore equal to approximately $30 \text{ protons} \cdot \text{mm}^{-2} \cdot \text{projection}^{-1}$. The image displayed in Fig. 3 contains the center slice (2.5 mm thick) of the reconstruction, with 180×180 pixels each 1 mm^2 .

E. Analysis Technique

The MTF was determined from the edge spread function (ESF) using a modified version of the oversampling method

Fig. 3. Reconstructed phantom using FBP as initial iterate and DROP with TVS. The binning size was 1 mm in t and 2.5 mm in v and the usual cuts were made at 3σ in angle and WEPL. The proton flux through a plane was approximately $28 \text{ protons} \cdot \text{mm}^{-2} \cdot \text{projection}^{-1}$. The image is 180×180 pixels with a pixel size of 1 mm^2 .

which was first described Judy [10] and later by Fujita et al [11]. In addition, modifications and improvements to the method described in papers by Mori and Machida [12] and by Watanabe et al. [13] were also used.

The two materials were juxtaposed to produce a sharp edge with a slight angle, α , 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 insure that all possible regions of the pixel can be sampled as the oversampling method requires. A 2-dimensional bilinear interpolation was applied to the reconstructed image to allow for subpixel-spaced sampling. Approximately 30 samples were selected, orthogonal to the edge, and the 50% values of each resulting ESF were overlaid to yield an oversampled ESF. For noise suppression, the oversampled ESF was rebinned using a bin size equal to the sampling pitch. A smoothing cubic spline was applied to the oversampled and rebinned ESF in order to optimize the χ^2 per degree of freedom value of the spline to the data. A line spread function (LSF) was obtained by differentiating the splined ESF. The MTF was obtained by Fourier-transforming the LSF.

III. RESULTS AND DISCUSSION

Fig. 4 illustrates the MTFs for each of the high contrast inserts, tooth enamel, cortical bone, lung (inhale) and air for each of 4 radial displacements (i.e. radial distance from the center of the phantom) $R = 27.5 \text{ mm}$, 42.5 mm , 52.5 mm and 67.5 mm . The low contrast inserts were not included in this study because they could not be resolved from the background due to low statistics. The intersections of the MTF with the magenta horizontal lines show the values of the $\text{MTF}_{10\%}$, the limiting resolution of the system. The observed $\text{MTF}_{10\%}$ values are between 0.25 and 0.35 lp/mm . The plots show that the limiting resolution tends to increase for increasing radial

(a) MTFs for Enamel Insert

(b) MTFs for Bone Insert

(c) MTFs for Lung Insert

(d) MTFs for Air Insert

Fig. 4. Plots illustrating the MTFs for 4(a) tooth enamel, 4(b) cortical bone, 4(c) lung (inhale) and 4(d) air for each of 4 radial displacements, $R = 27.5 \text{ mm}$, 42.5 mm , 52.5 mm and 67.5 mm . The intersections of the MTF with the magenta horizontal line shows the value of the $\text{MTF}_{10\%}$ which illustrates the limiting resolution of the system.

Fig. 5. Plot of limiting resolution versus radial displacement for various materials. As radial displacement increases, resolution of high RSP inserts increases significantly, but for low RSP inserts the resolution does not change significantly. The brain insert was not able to be resolved for all radial displacements.

displacement from the phantom center, in particular for the high RSP inserts (enamel and bone), while for the low RSP inserts the effect is much smaller (lung and air).

Fig. 5 shows the interplay between RSP and radial displacement in more detail. As radial displacement increases, resolution of high RSP inserts increases significantly, but for low RSP inserts the resolution first increases, then decreases again at the largest radial displacement. The results for high RSP inserts was expected: since for high RSP materials the MLP has the highest uncertainty at small radial displacement, we see a significant increase in spatial resolution as radial displacement increases from 27.5 mm to 67.5 mm for both enamel and cortical bone (3.38σ increase and 6.07σ increase respectively). The smaller increase for enamel can be accounted for because the MLP model assumes a uniform RSP in the phantom, which is not true when other material inserts are present. A greater non-uniformity along the path, caused by a higher RSP insert (enamel) results in lower spatial resolution.

The results for low RSP inserts provide evidence of a competing process: a distortion effect due to low density inhomogeneity. The MLP model appears to perform less accurately for low density materials such as air and lung, which cause little to no scattering. When these inserts are at the edge of the phantom, proximal to the beam, the MLP of the proton is incorrect at the very beginning of its trajectory and errors are compounded along its path through the phantom, causing an overall reduction in spatial resolution that worsens at the periphery. As a result, for low density inserts the radial displacement of the insert from the center of the phantom plays a less critical role.

The effect of bin size, independent of pixel size, on the limiting resolution of the system was also studied. Bin size determines the size of the distributions on which statistical cuts in angle and WEPL are made. It also provides the positional information required for the execution of the FDK algorithm. When binning size in t is reduced from 1 mm to 0.5 mm, the limiting resolution of the system was found to increase by 10-30%, for a pixel size of 1 mm^2 .

It is important to note that for a digital image reconstruction, the maximum expected resolution is the Nyquist frequency which is 0.5 lp/mm for a pixel of 1 mm^2 . Therefore, the spatial resolution we observed in this study accounts for 50 - 75% of the Nyquist frequency. The improvement achieved by reducing the binning size in t improves the resolution to 60 - 85% of the Nyquist frequency. This result suggests that the bin size in the FDK reconstruction is one of the limiting factors of image resolution in our system. By improving the initial image processing and reconstruction, we expect to achieve higher resolution in pCT.

IV. CONCLUSIONS AND FUTURE INVESTIGATION

We have showed that for high contrast inserts, the resolution of our system, given a binning size of $1 \text{ mm} \times 2.5 \text{ mm}$, and a pixel size of 1 mm^2 is about 50 - 75% of the Nyquist frequency. There is also evidence to suggest that by changing the binning size for the FDK we can further improve this result. We have also found that the MTF of inserts with high RSPs were more affected by radial displacement effects than inserts with small RSPs. Inserts with small RSPs display the effects of a competing process due to low-density inhomogeneity. Future investigations will focus on improving the MLP model and updating it after each iteration of the reconstruction to further improve spatial resolution in pCT.

The tools developed for this investigation can further be used to study the effects of various aspects of the system on image resolution, including the effects of MCS, non-ideal detectors, binning size and reconstruction algorithms. One of the next steps will be to apply this procedure to experimental data obtained from the phase II pCT scanner.

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