Design And Construction Of The 1st Proton CT Scanner

G. Coutrakon\textsuperscript{1}, V. Bashkirov\textsuperscript{2}, F. Hurley\textsuperscript{2}, R. Johnson\textsuperscript{3}, V. Rykalin\textsuperscript{1}, H. Sadrozinski\textsuperscript{3}, R. Schulte\textsuperscript{2}

\textsuperscript{1}Dept. of Physics, Northern Illinois Univ., DeKalb, IL 60115
\textsuperscript{2}Dept. of Radiation Medicine, Loma Linda Univ. Med. Ctr., Loma Linda, CA 92354
\textsuperscript{3}Santa Cruz Institute of Particle Physics, Univ. of California at Santa Cruz, CA 95064

Abstract: This paper discusses the design and operation of the 1\textsuperscript{st} proton CT scanner for 3D imaging. Reduction of proton range uncertainties and improved dose accuracy in the patient for treatment planning are central goals. A central CT slice acquired by reconstruction of 134 million proton tracks through a 14 cm spherical polystyrene phantom with high and low density inserts is presented.

Keywords: Proton treatment planning, proton range uncertainties, proton stopping powers, proton computed tomography

INTRODUCTION

Allan Cormack, Nobel laureate for X-ray Computed Tomography (CT), first proposed using proton CT for proton therapy[1] in 1976. The goal of 3D imaging with protons is to achieve better electron density resolution (or relative stopping power) and lower imaging dose than X-ray CT. In addition, high density artifacts should be eliminated, or at least greatly reduced. The result will be that proton range and proton dose distributions can be more accurately determined in proton therapy treatment planning. This in turn leads to tighter margins around the target volume and less dose to nearby critical organs. At present a 3\% distal margin is added to target volumes at most proton centers to insure proper dose coverage, largely due to range uncertainties attributable to systematic errors in converting X-ray Hounsfield units to relative proton stopping powers[ 2,3]. In this project we seek to reduce those proton range uncertainties to 1 or 2 mm; a significant improvement in range uncertainties over x-ray CT. These errors should be noticeably reduced when images are acquired from proton CT [9]. In this paper we present the design and beam test results for a prototype scanner capable of imaging a human head. As in x-ray CT, it is necessary to deliver protons at least 180 degrees around the patient and to record tracking information and energy loss through the patient for each proton. We anticipate that approximately one billion protons must be recorded and distributed through all gantry angles and in a time less than 10 minutes when the detector is fully developed. Since we wish to determine proton range to within +/- 1 mm for head, neck and brain tumors, the voxel size for spatial resolution should be of order 1 mm for the image reconstruction, similar to X-ray CT images. We estimate that a proton beam of 200 MeV entering the patient will be adequate for complete transmission through a human head.

General Theory of Proton CT

When protons lose energy through the patient, they exit the body with an energy that is reduced by \(\Delta E\). The relative stopping power (RSP) for each voxel traversed is related to the exiting energy, \(E(\text{out})\) by

\[ \int_{\text{path}} \text{RSP}(l)dl = \int_{E(\text{in})}^{E(\text{out})} \frac{dE}{S(\text{water})} = WEPL \]  

(1)

where \(E(\text{out}) = E(\text{in}) - \Delta E\), \(S\) is the mass stopping power for protons in water and \(WEPL\) is the water equivalent path length for each track through the phantom. "Path" means the most likely path length (MLP) through the body and \(E(\text{in})\) is a constant of order 200 MeV. The integral in Eq. 1 can be approximated by a discreet sum of RSPs times the chord length for each voxel crossed by the track. This sum is the WEPL for that track. Writing this linear
equation for each track leads to a system of linear equations that can be used to solve for the RSP for each of the voxels. In matrix notation this becomes

\[ \mathbf{A}(\text{RSP}) = (\text{WEPL}) \]  \hspace{1cm} (2)

where \( \text{WEPL} \) is the \( n \) dimensional vector of water equivalent path length for \( n \) protons through the anatomical region of interest. \( \text{RSP} \) is an \( m \) dimensional vector which represents the solution set for the \( m \) voxels in the region of interest. The proton CT reconstruction problem reduces to solving this system of \( n \) equations and \( m \) unknowns with \( n > m \). The matrix \( \mathbf{A} \) corresponds to the chord length of each track through each voxel. Since only a few hundred voxels are crossed by any given track, the elements of matrix \( \mathbf{A} \) will be mostly 0's, and hence, a quite sparse, but nonetheless, quite large matrix; one with potentially \( 10^5 \) columns and \( 10^9 \) rows for an adult head size phantom. The goal of the tracker is to measure the entrance position and angle of each track while the calorimeter measures the residual energy, \( E(\text{out}) \), to determine the \( \text{WEPL} \) for each track. The entrance and exit trajectories allow one to construct the most likely path (MLP) of each proton through the patient and hence the chord length through each voxel. The calorimeter measures the energy after the patient and is used to evaluate the integral on the left side of Eq. 1, namely, the water equivalent path length through the patient.

\[ \text{Figure 1.} \text{ The proton CT detector concept with head phantom on rotational stage. Four (x,y) tracker planes and a CsI calorimeter are used for the scanner.} \]

\[ \text{Design Considerations} \]

A conceptual design of a proton CT scanner for a head size phantom is shown in Fig. 1. Two X, Y planes of detectors, called trackers, measure the position at two Z locations both before and after the patient. This allows determination of both location and angle of the proton as it enters and exits the patient which in turn allows one to calculate the most likely path (MLP) through the patient. The calorimeter measures the residual energy exiting the patient and determines the vector (WEPL) on the left side of Eq. 2.

The goal of this first generation pCT scanner will be to image a head size phantom of 16 cm diameter and 16 cm along the body axis. Achieving 1% density resolution in each 1 mm\(^3\) voxel requires a ratio of particle tracks to number of voxels to be 100:1 [4]. Thus, with 3 million voxels, we expect 300 million tracks will be needed for reconstruction. The maximum particle data rate of this 1st prototype is 180 kHz which will require a two hour scan time using the LLUMC synchrotron with 25% duty factor for extracted beam. The next generation detector, currently being designed, will boost this rate to 2 MHz so that faster scan times can be achieved. In addition, the duty factor of the LLUMC synchrotron will be increased to 75% by using a longer beam spill.

\[ \text{Figure 2.} \text{ Data flow and compression from SSD (Silicon strip detector) to PC} \]

The lateral dimensions of the scanner determine the maximum field of view. Cost constraints limited the active area of this prototype to be 9 x 18 cm. Scanning a head with 16 cm along the body axis will require two consecutive scans and the images will need to be fused for the full field of view.

\[ \text{The Tracker} \]

The tracking detector consists of 8 planes of 400 \( \mu \)m thick silicon wafers (or sensors) with implanted strips for X or Y coordinate measurements. The multiple coulomb scattering through 16 cm of water like material will be several mm for 200 MeV protons. Simulations have shown [5] that with perfect position resolution, the path reconstruction errors are slightly less than 1 mm (rms) and that detector resolution up to 1 mm pitch will not significantly change this. Therefore, we have chosen a Si strip pitch resolution of 228 um which has been used previously for the GLAST (Gamma ray Large Area Space Telescope)
built by SCIPP at UCSC [6]. This also allowed us to use an existing design for the 6000 channels of close packed electronics after some minor changes. The maximum size of Si wafers that are commercially available is only 9 x 9 cm when cut to a square size. In order to achieve a 9 x 18 cm area, two wafers are placed together with a 2 mm overlap to avoid dead area. This effectively doubles the number of readout channels in one dimension so that each of the 8 planes has 768 channels for the 9 x 18 cm area. The Si strips from each plane are individually wire bonded to 6 ASICs (application specific integrated circuits), each containing 64 channels placed on a PC board. Data flow through the electronic stages is shown in Fig. 2. The front end 64 channel ASIC is the GTFE (GLAST tracker front end) which contains the preamplifier, pulse shaper and discriminator to register events with signal above threshold for each channel. The GTFE preamplifier is a charge sensitive amplifier with 1500 electron noise (rms) followed by a discriminator which provides a binary output to find “hit” strips. The pulse width and discriminator output is less than 10 µsec wide which sets the resolving time of the tracker for two successive events. The mean signal for each proton track of 200 MeV is approximately 25,000 electrons on the Silicon strip; well above the 1500 electron noise of the preamplifier. Six GTFE chips are used to read out each group of 384 channels. The GTRC (GLAST tracker Readout Controller) interfaces with the six GTFEs and compresses the data so that only the addresses of the hit channels are transferred back to the FPGA. The FGGA reads out all 16 GTRCs in parallel. The data is then passed to the host computer via Ethernet. The FPGA has an event buffer capable of storing up to 500,000 events before transferring to the host PC.

The Calorimeter

The calorimeter should have an energy resolution which does not exceed the energy straggling in the patient which is of order 1%; close to the best achievable resolution for modern calorimeters. Because cost was again a limitation, we chose 18 (3 x 3 cm) CsI crystals for the energy measurements in order to be well matched to the silicon strip detector area of 9 x 18 cm. Each of the 18 crystals were carefully wrapped in 65 µm reflective polymer film and later, packed closely together to minimize the dead space between the crystals. The crystals were fabricated with rough-cut surfaces except for the face coupled to the photodiode. This surface cut will degrade light output of the crystal, but will result in a more uniform response along the length of the crystal. The crystals measure 125 mm in length, sufficient to stop 200 MeV protons. To avoid deterioration of the photodiode surface, we used a silicone rubber film of 1.5 mm thickness as a coupling medium. For photo receivers, we used a Hamamatsu (HPK) S3584 photodiode which has a quantum efficiency well matched to the light emission spectra of CsI. The front end electronics for each of the 18 CsI crystals is composed of a CREMAT 110 [7] preamplifier and a CREMAT 200 pulse shaper. The noise of the fully depleted detector and preamplifier system is less than 3000 electrons (rms). For digitization, we use a 12 bit ADC, the least significant bit of which determines the limit of energy resolution of the front end electronics and ADC system. The ADC is readout by a Field Programmable Gate Array (FPGA). A fast trigger for data acquisition can be created by splitting the signal after the preamplifier for each crystal and summing the 18 outputs together.(1)

Data Acquisition System

A schematic of the data flow for the tracker is shown in Fig. 2 and integration with the calorimeter readout is shown in Fig. 3. A Xilinx ML507 development board with a Virtex 5 FPGA and PowerPC embedded CPU is used for the readout. The FPGA interfaces directly with the calorimeter ADC, the tracker readout ASICs, and triggering system. A host PC is responsible for recording the data and setting configuration options.

The PC runs software written in C++ and uses ROOT for data processing and storage. The PC uses an ethernet connection at 1 Gbps to connect to the FPGA. Data rate of 150,000 events per second has been tested, but the average data rate for the 1st pCT scans was closer to 50,000 events per second, due to the non uniform nature of the spill structure and random events occurring closer in time due to Poisson statistics.

Our choice of DAQ implementation easily permits changes in the trigger configuration as well as data flow type. In the current configuration there are two trigger options: one trigger can be created in the tracker FPGA by requiring a coincidence of one or more hits in one or more (x,y) planes; the other from
the fast calorimeter trigger signal generated by a sum of the 18 CsI crystal signals.

Test Configuration and Results for the 1st Images with a Spherical Polystyrene Phantom

![Diagram of experimental setup](image)

**FIGURE 4.** Fig. 4 Experimental layout of the cone beam proton CT system

Fig. 4 shows the experimental arrangement of the first beam tests with the full scanner and a 14 cm diameter phantom (a Lucy phantom from Standard Imaging, Inc) mounted on a rotational stage. The actual scanner with a phantom is shown in Fig. 5.

![Image of scanner with phantom](image)

**FIGURE 5.** The full scanner with rotatable head phantom on a mobile cart

A 200 MeV proton beam passed through 2 mm Pb foil located 2 meters upstream of the Lucy phantom to spread the beam. The phantom had four, 1 cm diameter, cylindrical inserts at the corners of a 6 cm square in the upper hemisphere to test data reconstruction of the different densities. 140 million tracks were recorded over 360 degrees (4 degree increments) for the reconstruction before cuts were made.

Image reconstruction software written by Scott Penfold [8] was used to reconstruct the images with 2.5 mm thick slices as shown in Fig. 6. The Lucy phantom is composed primarily of polystyrene and the four inserts are: 1) air, upper left, 2) Bone, lower right, 3) polystyrene, lower left with 2 mm aluminium beads (not yet visible), and 4) Lucite, upper right with 2 mm oil beads (also not yet visible). The reconstructed Relative Stopping Powers for polystyrene, bone substitute, acrylic, and air were 1.035, 1.68, 1.19, and 0.05. With the exception of air which contributes negligibly to a proton range error, the errors in these mean values were within 1% of the calculated values.

![Image of CT slice reconstruction](image)

**FIGURE 6.** First CT slice reconstruction of a 3D heterogeneous polystyrene phantom with air, bone, and acrylic inserts.

CONCLUSIONS

The first pCT scanner has been tested at the James M Slater Proton Treatment and Research Center in Loma Linda, California. Characterizations of tracker and calorimeter responses to mono energetic beams have been performed and the data acquisition system has been described. The first pCT images using a Lucy QA phantom from Standard Imaging have been generated from proton beam data. Initial results are encouraging but further studies of spatial and density resolution are needed before it can be clinically useful. Further tests are needed to determine the present limitations in reconstruction accuracy. This may include more dose for better statistical accuracy and/or better linear solvers for the matrix equation, Eq. 2. More simulations and pCT scans will help to investigate these questions. Additionally, a new detector design based on plastic scintillators for the calorimeter and tracking planes is now in progress to speed up data acquisition to obtain human head size images in under ten minutes for the next generation pCT scanner along with a larger imaging area (27 x 36 cm).

ACKNOWLEDGEMENTS

The authors wish to thank the US Department of Defense (DOD), US Army Medical Research Activities and Acquisitions, Ft. Detrick, MD, for
sponsoring this project through Northern Illinois University and Loma Linda University Medical Center. The support of Dr. James M. Slater (LLUMC), Dr. John Lewis (NIU), and Ms. Kathy Buettner (NIU) has made this project possible through their strong encouragement and communications with the principle investigators and the DOD funding agency. Finally, we thank Dr. Scott Penfold who performed the image reconstruction (Fig. 6) with software written for his PhD thesis. These results would not have been possible without his enormous effort to the project.

REFERENCES

6. www.scipp.ucsc.edu/groups/glast/publications