Reconstruction for Proton Computed Tomography: A Practical Approach

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***Abstract*- Proton Computed Tomography (CT) has important implications for both image-guided diagnosis and radiation therapy. For diagnosis, the fact that the patient dose committed by proton CT compares favorably with that delivered by traditional X-ray CT, for the same density resolution and contrast, may be exploited in dose-critical clinical settings. Proton CT is also the most appropriate imaging method to perform planning and verification of proton-based radiation treatment, since proton stopping power maps constructed by table-based transformation of X-ray CT images only render approximate stopping power estimates.**

In proton CT, sharp features become blurred by the phenomenon of multiple Coulomb scattering (MCS), resulting in a resolution of around 3 to 5 mm. Studies showed that the spatial resolution of proton radiography and CT can be improved to about 1-2 mm by tracking *individual* protons in coincidence as they enter and exit the imaged object. This paper describes a new practical implementation of this approach. We first bin the captured protons into slots of similar tracks. Optionally, proton energy statistics can be collected within each bin to obtain further parameters for tissue characterization. The envelope of path uncertainty due to MCS can be be modeled as a banana-shaped curve. The 3D reconstruction, using either filter-backprojection or iterative algorithms, can be performed rapidly on graphics hardware, using a slice blurring technique to model the MCS uncertainty curve.

I. Introduction

PROTON Computed Tomography (CT) has important implications for both image-guided diagnosis and radiation therapy. For diagnosis, the fact that the patient dose committed by proton CT compares favorably with that delivered by traditional X-ray CT, for the same density resolution and contrast, may be exploited in a number of clinical settings, e.g., in cases in which a patient needs to be imaged frequently or in sensitive tissue areas. Proton CT is also the most appropriate imaging method to perform planning and verification of proton-based radiation treatment, since proton stopping power maps constructed by table-based transformation of X-ray CT images only render approximate stopping power estimates, limiting the power of proton radiation therapy. This paper discusses the fundamental aspects of proton CT, starting with the current state of affairs and then proceeding to an overview of our new proposed approach. A companion paper, also submitted to this conference, will give more detailed renditions on a preliminary study using a Monte Carlo simulated dataset [3].

II. Background

An increasing number of cancer patients is treated with proton radiation worldwide. Proton radiation therapy is one of the most precise forms of image-guided cancer therapy since the sharp dose peak (called the *Bragg peak*) produced by protons at the end of their range can be placed anywhere inside the patient's body, and can deliver a destructive dose to the target while completely sparing tissues behind the target. An important aspect of proton therapy is the accuracy of the range determination of protons in tissues. When sensitive structures are in close proximity to the tumor the most appealing approach of stopping the proton beam immediately in front of the critical structure is often avoided by the radiation oncologist due to proton range uncertainties and the fear of side effects. What is needed is an accurate map of the proton stopping power in the traversed tissue such that the Bragg peak can be delivered precisely to the tumor site. Proton stopping power is linearly related to the electron density of tissue, while X-ray CT is dependent on attenuation coefficients of the tissues for a spectrum of photons. Due to these fundamental physical differences, the accuracy of mapping X-ray Hounsfield numbers to proton stopping powers cannot be better than a factor of 3%, and range uncertainties can exceed 10 mm in a human head [12].

Cormack was probably the first to point to the possibility to perform proton CT, in 1963 [3]. In early studies proton radiographs were taken in a fashion very similar to x-ray radiographs: a passive photographic plate behind the sample recorded projected images of an incident beam, according to the transmission efficiency of the protons [8]. The proton beam energy had to be tuned to center the Bragg peak on the photographic plate in order to get useful density contrast. With

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this method, sharp features become blurred by the phenomenon of multiple Coulomb scattering (MCS), resulting in a resolution of around 3 to 5 mm. Later studies showed that the spatial resolution of proton radiography and CT can be improved to about 1-2 mm by tracking *individual* protons in coincidence as they enter and exit the imaged object [6][13]. Our implementation uses this most accurate approach.

III. A PRACTICAL RECONSTRUCTION APPROACH

We shall assume that the incident angle (a pair in 3D) and the energy of the individual protons is available at good accuracy (see Fig. 1). A description of the equipment with which this can be achieved is subject of another paper [14] With regards to the 3D reconstruction, there are a number of challenges that need to be addressed:

- the large angular distribution of the particles must be accommodated
- the scattering envelope of the particles must be modeled
- the massive amount of projection data must be managed efficiently
- the reconstruction must be accomplished within a reasonable amount of time, i.e., on the order of minutes

The last item on this list determines the type of approach we feel is required to faciliate a practical implementation. We accomplish the first and third task on the list by sorting the data into bins $(\alpha, \beta, \mathbf{s})$, where $\mathbf{s}$ is the location in the sinogram and (α, β) are the (discretized) horizontal and vertical angles of ray inclination with respect to the flat detector plane. The energy distribution within each bin can be used to compute probalistic measures for characterizing the object tissues encountered along the particle paths. We shall use the average energy for now. We then store the (image) data sorted by (α, β).

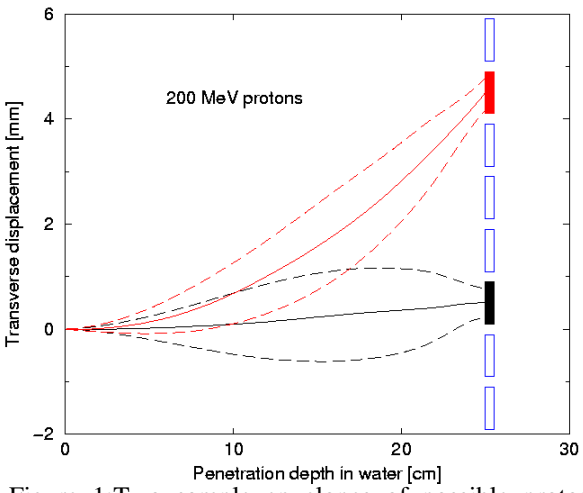


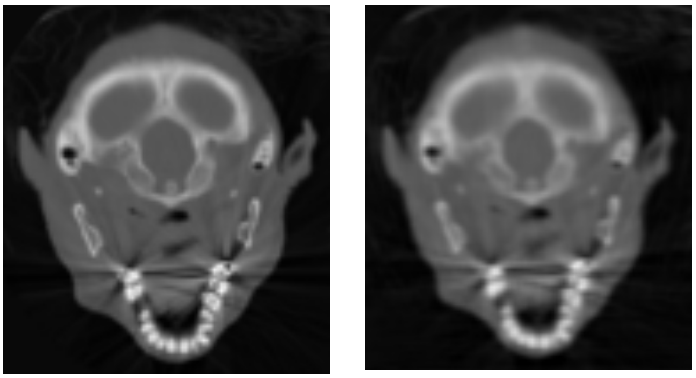
Figure 1:Two sample envelopes of possible proton trajectories due to multiple Coulomb scattering

For reconstruction, we may either use a Filtered Backprojection method, such as the one by Feldkamp et al. [4], or an iterative method, such as MLE [11] or SART [1][5]. The former is generally faster, but does not deal well with incomplete projection data. It may be the case that some angles or bins are not selected by particles during an imaging session, which will give rise to undesirable reconstruction artifacts caused by the initial filtering stage. In these cases, we may resort to (slower) iterative methods that generally perform better in these scenarios. With respect to the significant Coulomb scattering, MLE is also an appropriate choice to maximize the fit of the reconstruction to the data. We suggest to use Ordered Subsets-EM (OS-EM) [7] for faster convergence.

Clinical utility requires fast reconstruction within minutes. This is best afforded by performing the reconstruction exploiting commodity graphics hardware boards (GPUs), which are currently evolving in their capabilities. A reconstruction is performed by using the texture mapping facilities of the hardware for performing the interpolations required in the projection and backprojection operations [2][10]. The other arithmetic operations that occur in the reconstruction algorithms can also be performed in this hardware.

Finally, the probablity envelope can be modeled by performing a slice blurring before the projection summing or volume update during projection and backprojection, respectively. We determine the amount of blurring as a function of the width of the envelope at the slice position.

Using our hardware reconstructor we were able to perform a cone-beam reconstruction (16° cone-angle) of a 128^3 volume from 80 projections in 3 SART iterations in 50 seconds, which is clearly in the clinical range. Fig. 2 shows a slice of a reconstructed CT head (b) and the original (a).



(a) **(b)**

Figure 2: CT head dataset: (a) original, (b) reconstructed from 80 high-quality simulated projections acquired from (a).

IV. CONCLUSIONS

The approach we have outlined here forms the general data processing pipeline for a proton CT application, past the projection data pre-processing stage. As we experiment more with both Monte-Carlo simulated as well as real data, we hope to come up with effective and accurate models for the probability envelope as a function of (α, β). We also hope to

incorporate some simple Monte-Carlo simulation capabilities directly into our hardware reconstruction platform, in order to better model the Coulomb scattering processes during the 3D reconstruction.

V. DISCLAIMER

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