A Proton Computed Tomography Demonstrator for Stopping Power Measurements

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Motivation – Particle therapy

Advantages of ion-beam therapy over photon therapy

- Energy deposition (dose) in ion-beam therapy strongly localised \( S \propto \frac{1}{v^2} \)
- Accurate dose-deposition
- Treatment of tumors close to radio-sensitive tissues, e.g. optical nerve

Photon therapy:
\[ I = I_0 e^{-\mu x} \]

Ion-beam therapy:
\[ \bar{R}(E_0) = \int_{E_0}^{0} \frac{1}{S(E)} dE \]
with \( S(E) = -\frac{dE}{dx} \)

Image: Dose comparison for photon (left) and proton (right) treatment plans. (Linz 2016)
Motivation – Treatment planning

Treatment planning based on X-ray CT

- Conversion errors from Hounsfield units (HU) to relative stopping power (RSP) lead to range errors ($\approx 1 - 3$ mm) (Schaffner et al. 1998)

$$HU = 1000 \times \frac{\mu - \mu_{water}}{\mu_{water}}$$

$$\Downarrow$$

$$RSP = \frac{S(E)}{S(E)_{water}}$$

- Solution: direct measurement of stopping power (imaging with ions)
  - Same particle type for treatment planning and therapy

Image: Conversion from HU to RSP (Schaffner et al. 1998)
Imaging with ion beams – Overview

Particles with energy $E$:
- Pass front tracker
- Lose energy in object: $\Delta E$
- Pass rear tracker
- Deposit energy in calorimeter: $E - \Delta E$

Proton/ion CT:
- Calculate path estimate and measure $\Delta E$ for each ion
- Repeat for different object rotations
- Reconstruct 3D image of stopping power
pCT setup – Overview

- pCT demonstrator based on double sided silicon strip detectors (DSSDs) and a range telescope
- Synchronisation via AIDA2020 trigger and logic unit (TLU) (Cussans 2017)
  - Exclusive trigger number per particle to correlate tracks and energy loss
- Object to be imaged (1 cm³ aluminum cube phantom) was mounted on a rotating table
pCT setup – Tracker

Tracking telescope with 2+2 DSSDs

- **DSSD**
  - Thickness: 300 µm
  - Size: \((2.56 \times 5.12)\) cm\(^2\)
    - X-side: 512 p-doped strips with 50 µm pitch
    - Y-side: 512 n-doped strips with 100 µm pitch

- **GbE-based readout**
  - APV25 chip (French et al. 2001)
  - Belle-II SVD readout chain with adapted FW and SW (Thalmeier et al. 2017)
  - Achieved event-rate \(\leq 30\) kHz

- **Corryvreckan framework for tracking** (Dannheim et al. 2021)
  - Detector alignment
  - Track fitting
  - Phantom positioning based on MCS radiography (Schütze et al. 2018)
Implementation of a range telescope (formerly TERA (Bucciantonio et al. 2013))

- 42 plastic scintillators layers
  - Size: \((3 \times 300 \times 300) \text{ mm}^3\) \((\approx 3.6 \text{ mm WET})\)
  - Can measure protons up to \(\approx 150 \text{ MeV}\)

- SiPMs for signal generation
  - 400 pixel
  - Subsequent ADC resolution (12 bit)
  - Limited energy range
    - Range telescope instead of sampling calorimeter

- Readout via USB connection
  - Event rate < 16 kHz

- SiPM power supply was unstable
  - Mainboard was completely redesigned
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pCT setup – Calorimeter calibration

➢ Optimization of SiPM dynamic range
   ▶ SiPM calibration was performed with 800 MeV protons at MedAustron

➢ Calibration of range telescope
   ▶ Estimation of mean water equivalent thickness (WET) of the calorimeter components
      ⊳ Ranges are measured for different proton energies
      ⊳ Comparison to NIST data for WET estimation of trigger scintillators and TERA scintillators

➢ Range algorithm for single protons
   ▶ Energy cuts in plateau (pile-ups)
   ▶ Last slice over threshold and first slice under threshold defines range
      ⊳ To compensate fluctuations of single slices

Fit proton beam to NIST: \( R_{\text{WET}}(E) = WET_{\text{offset}} + WET_{\text{slice}} \cdot R_{\text{slice}}(E) \)

\begin{align*}
\text{WET}_{\text{offset}} &= 36.61 \pm 1.04 \text{ [mm]} \\
\text{WET}_{\text{slice}} &= 3.56 \pm 0.06 \text{ [mm]}
\end{align*}
pCT setup – Testbeam at MedAustron

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Results – 2D Projections

Performed measurements

- 100.4 MeV protons
- 77 projections with $\approx 2.5 \times 10^6$ events (24 min) each
- However, only $\approx 6.5 \times 10^5$ synchronized events per projection (mean event rate $\approx 450$ Hz)

  - After tracking, synchronization and cuts

![Forward projection (720346 entries)](image1)
![Forward projection (579525 entries)](image2)
![Forward projection (643560 entries)](image3)
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Results – 3D Reconstruction

Reconstruction

➤ Tomographic iterative GPU-based reconstruction toolkit (TIGRE) (Biguri et al. 2016, Kaser et al. 2021)
  ▶ Limited amount of projections
  ▶ Straight-line approximation

Image: 3D view of reconstructed aluminum phantom

Performance

➤ RSP values in region of interest were summarized and compared to literature values

RSP resolution: \( \frac{\sigma_{\text{RSP}}}{\mu_{\text{RSP}}} \approx 6.4\% \)

RSP accuracy: \( \frac{\mu_{\text{RSP}} - \text{RSP}_{\text{lit}}}{\text{RSP}_{\text{lit}}} \approx 8.2\% \)
Discussion – Current Limitations and requirements

Current demonstrator was build from existing hardware

- Full pCT workflow was established
- Not designed for clinical use
- New demonstrator design is studied

Requirements for a clinical pCT system (Schulte et al. 2004, Bashkirov et al. 2016)

- Event rate: > 1 MHz
- Spatial resolution (image): < 1 mm
- Energy resolution: < 1 %
- RSP accuracy: < 1 %

Upgrade solution: 4D tracking detectors

Image: TOF calorimeter for pCT

\[
E_{kin} = m_0c^2 \cdot \left( \frac{1}{\sqrt{1 - \frac{L^2}{c^2TOF^2}}} - 1 \right)
\]
Upgrade solution – 4D tracking detectors

Design studies for potential upgrade solution based on MC Simulations

- Full pCT system based on one detector technology ⇒ LGADs
- System parameters were optimized to fulfill clinical requirements
- State-of-the-art reconstruction algorithm (distance-driven binning (Rit et al. 2012))

- CTP404 phantom was used for performance studies
  - PMMA cylinder (15 cm diameter)
  - Inserts with different materials
  - RSP accuracy and precision was measured in inserts
RSP precision

- Energy resolution and RSP resolution are mainly dominated by intrinsic time resolution per plane and beam energy.
- Intrinsic TOF resolution should be $\leq 30$ ps per plane to fulfill clinical requirements ($E_{\text{res}} < 1\%$).
RSP accuracy

- Dedicated calibration procedure for TOF calorimeter was implemented
- After calibration RSP accuracy could be lowered down to $\approx 0.15$-$0.5\%$
  - Well below clinical requirements

![Graph 1](image1.png)

Mean absolute percentage error of RSP for 1m flightpath and 0.1% X/X0 LGADs

![Graph 2](image2.png)

Mean absolute percentage error of RSP for 1m flightpath and 2.3% X/X0 LGADs

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Summary

- Ion imaging can potentially improve treatment accuracy
- A working pCT demonstrator was constructed
  - (Partly damaged) range telescope had to be repaired and upgraded to get a functional pCT demonstrator
- A 3D image of an aluminum phantom was recorded at MedAustron
- Design studies for a possible upgrade solution are ongoing
  - A pCT setup based on 4D tracking detectors could potentially fulfill clinical requirements

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Accelerator layout
Accelerator layout – Synchrotron

- Circumference: 78 m
- Radius: 12 m
- 16 dipole magnets
- 24 quadrupole magnets
- 1 RF cavity for acceleration

Image: MedAustron
Backup— MedAustron

Synchrotron accelerator complex

- Circumference: 77.4 m
- Energies:
  - Protons: 60 MeV to 800 MeV, Clinical energies $\leq$ 250 MeV
  - Carbon ions: 120 MeV/u to 400 MeV/u
- 4 slots for ion sources:
  - Protons
  - Carbon ions
  - Redundant source
  - Unused, could be used for He
Backup– MedAustron

Synchrotron accelerator complex

- Four irradiation rooms:
  - **IR1**: Exclusive to research
    (protons up to 800 MeV, low rates)
  - **IR2, IR3, IR4**: Clinical use
    (Limited to clinical energies)
  - Beam only in one room at a time

- Beam parameters:
  - Beam delivery: pencil beam scanning
  - 5 s spill
  - Spotsize: 7 mm to 21 mm FWHM
  - Clinical rates:
    - Protons: $10^9$ particles/s
    - Carbon ions: $10^7$ particles/s
  - Research: $\geq 10^3$ particles/s
Backup – Reconstruction

Imaging in a nutshell:

- **Forward projection** $p_i$ (Radon transform):
  \[
  R[f(x, y)] = \int_{\gamma(\alpha, r)} f(x, y) \, ds \equiv p
  \]

- Insert physics to define forward projection:
  \[
  \int SPR(x, y) \, dl = b(E_{in}, E_{out})
  \]
  \[
  p_i = b(E_{in}, E_{out})_i \approx \sum w_{i,j} SPR(x, y)_j
  \]

- Forward projection is a set of linear equations
  \[Ax = b, \text{ with } b_i \text{ as a function of the residual energy of particle } i, \ x_j \text{ as the SPR in voxel } j \text{ and } A_{i,j} \text{ as the particle's pathlength through voxel } j\]

- **Backprojection** means solving linear equations
  \[Ax = b \Rightarrow x = SP\]
Backup – TIGRE toolbox

- **TIGRE: Tomographic Iterative GPU-based Reconstruction Toolbox**
  - Developed for cone beam CT (CBCT)
    - Used by collaborating group at MedUni Vienna for CBCT
  - Single or multi-GPU computation
  - Modular structure
  - Forward and backprojection ($A(x)$) are optimized for GPU computing
  - Algorithms are written in high-level language (Python, Matlab)

Available algorithms:
- Filtered back projection, FDK
- Iterative algorithms (SART, OS-SART,..)
- Custom algorithms

Image: TIGRE (Biguri et al. 2016)

7 https://arxiv.org/abs/1905.03748
Backup – Tracker readout system

6 DSSD modules:
2 x 4 APV25 chips, 128 channels each

2 Junction boards:
rad-hard DC/DC converters

2 FADC boards:
signal digitization, zero supression and VME interface

EPICS based slow and run control:
data readout via VME bus

Front-end power supply

~5m twisted pair cable

VME bus

EPICS

LV

HV
Use only tracker clusters to radiograph a scattering body
   ▶ No energy measurement

Fit tracks separately for tracker triplets before and after scatterer

Calculate angle between tracks
   ▶ x- and y-direction:
     two independent measurements

In each bin:
   1 Collect distribution of scattering angles for intercepting tracks
   2 Calculate width (variance) of the distribution of angles
Backup – Multiple scattering radiography

- Scattering angle distribution is centered around zero
- Its width depends on the integrated material budget $\varepsilon$ that particles pass
  - $\varepsilon = x/X_0$
- Using the Highland formula, $\varepsilon$ can be reconstructed:
  $$\Theta^2(L) \approx \left( \frac{13.6 \text{ MeV}}{\beta c p} \cdot z \right)^2 \int_L \frac{1}{X_0(x,y,z)} |ds|$$

Image: Scattering based radiography of the stair phantom.
SiPM operates in Geiger mode
- Light from scintillators is measured with SiPMs
- Signal is proportional to number of detected photons (fired pixels) → $E_{\text{dep}}$ in scintillator

Limited energy range and resolution
- Only 400 pixel
- Subsequent ADC resolution (12 bit)

Energy deposition in scintillator $\propto$ Landau distribution
- MPV is shifted by adapting bias voltage (gain) to optimise energy range and energy resolution
  * Gain very sensitive to voltage instabilities and temperature
- MPV of ADC counts is then converted to deposited energy by comparison to Geant4 simulation (calibration)
Backup – TOF CT analysis

Figure: a gauss fit to RSP distribution in each insert was applied

- the RSP values were collected in $7 \times 7 \text{mm}^2$ squares at the centre of each insert.
- 15 slices per insert were used for the RSP distribution
- the relative RSP error, the mean absolute percentage error (MAPE) and coefficient of variation were obtained to measure the performance of the CT setup

$$CV = \frac{\sigma_{\text{RSP}}}{\mu_{\text{RSP}}} , \quad err_{\text{rel}} = \frac{|\text{RSP}_{\text{theo}} - \text{RSP}_{\text{meas}}|}{\text{RSP}_{\text{theo}}}$$

$$\text{MAPE} = \frac{\sum_i n_{\text{mat}} err_{\text{rel},i}}{n_{\text{mat}}}$$


