

Characterization of solid-state detectors for proton beam dosimetry using Monte Carlo simulation techniques



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Introduction

Proton therapy

- + Highly localized dose delivery
- + Less damage to surrounding tissue
- + Fewer side effects
- Difficult dose validation
- Higher need for spatial assurance



Need for alternatives for QA/QC & patient dosimetry

Luminescent detectors

- Broadly used for dosimetry (personal, medical, ...)
- Adjustable size
- Adjustable shape

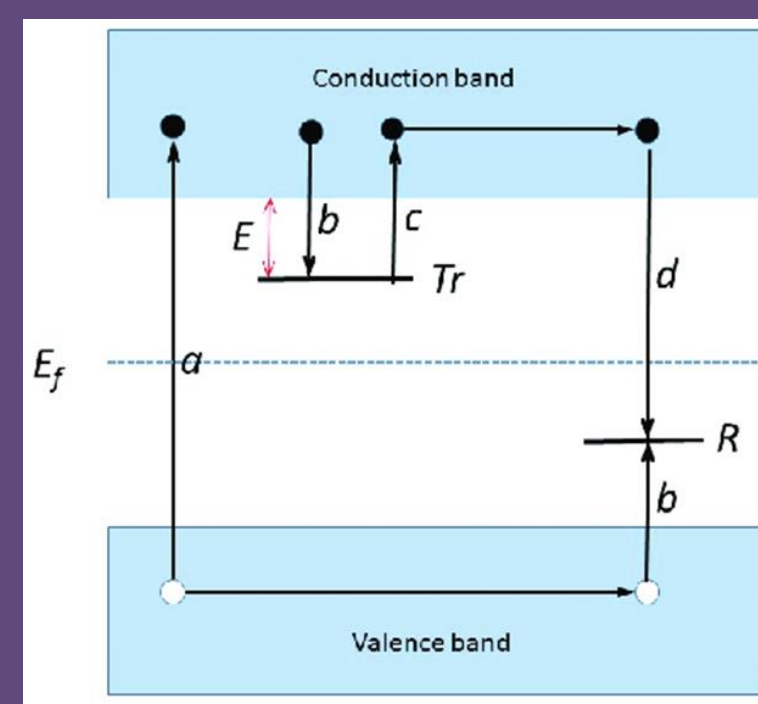


Figure 1: Principle of luminescent crystals.

Radiation dosimetry

Absorbed dose

$$D = \frac{d\varepsilon}{dm}$$

- $d\varepsilon$: deposited energy
- dm : the mass

Stopping power

$$S = -\frac{dE}{dl}$$

- dE : energy loss
- dl : travelled length

Materials & Methods

Proton Beam

Energy:

- ✓ 60 MeV (intraocular tumors)
- ✓ 230 MeV (deep tumors)

Beam characteristics:

- ✓ Conical pencil beam
- ✓ Placed 1m above the phantom
- ✓ Angular divergence $\vartheta = 0.14^\circ$
- $\varphi = 360^\circ$

Simulation Geometry

Phantom

- ✓ Material: H₂O
- ✓ Cylinder
- ✓ Radius 5 cm
- ✓ Length variable (due to different energies)

Dosimeters

- Materials:
 - Al₂O₃
 - BeO
 - CaSO₄
 - H₂O (reference)
- Cylinder
- Radius 1 mm
- Length 1mm

PENELOPE/PENH: Monte Carlo transport Code

- ✓ Extension PENH to include proton beams
- ✓ Mainly for dosimetric and clinical applications
- ✓ Nuclear reactions of alphas and deuterons are computed as if they were protons

Results & Discussion

Fitting with Crystal Ball function:

$$CB(z, z_0, \sigma, a, n, N) = \begin{cases} \exp\left(-\frac{(z-z_0)^2}{2\sigma^2}\right) & \text{for } \frac{z-z_0}{\sigma} > -a \\ N \cdot \left[\left(\frac{n}{|a|}\right)^2 \cdot \exp\left(-\frac{|a|^2}{2}\right) \cdot \left[\frac{n}{|a|} - |a| - \frac{z-z_0}{\sigma}\right]^{-n}\right] & \text{for } \frac{z-z_0}{\sigma} \leq -a \end{cases}$$

N: normalization factor, z_0 : peak's depth, σ : standard deviation, n : order of power tail, a : boundary of the two distributions.

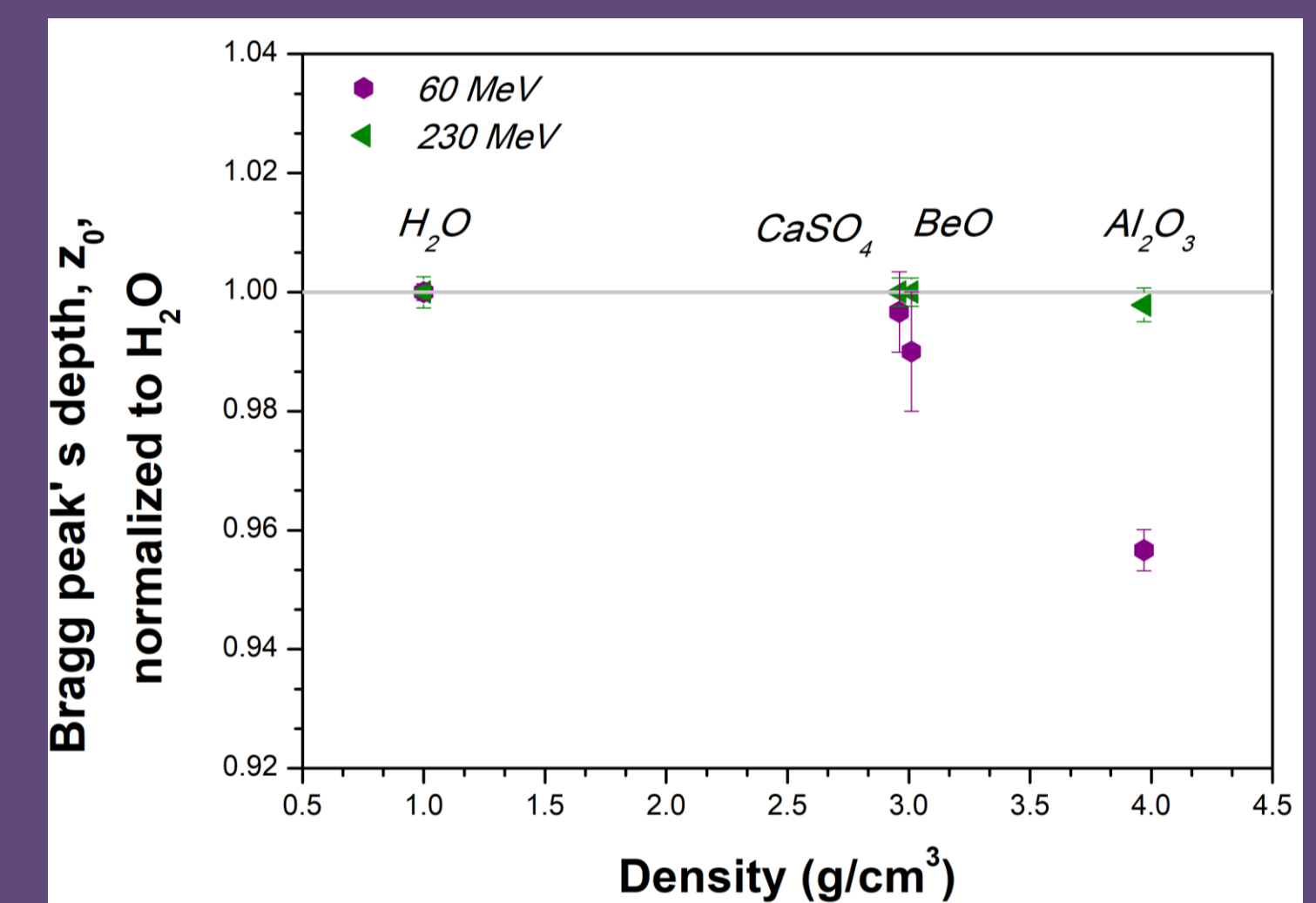


Figure 5: Location of Bragg peak with respect to the dosimeter's density.

Track averaged LET

$$\varphi - LET(z) = \frac{\sum_{i=1}^N \phi_i(z) S_i}{\sum_{i=1}^N \phi_i(z)}$$

Dose averaged LET

$$D-LET(z) = \frac{\sum_{i=1}^N \phi_i(z) S_i^2}{\sum_{i=1}^N \phi_i(z) S_i}$$

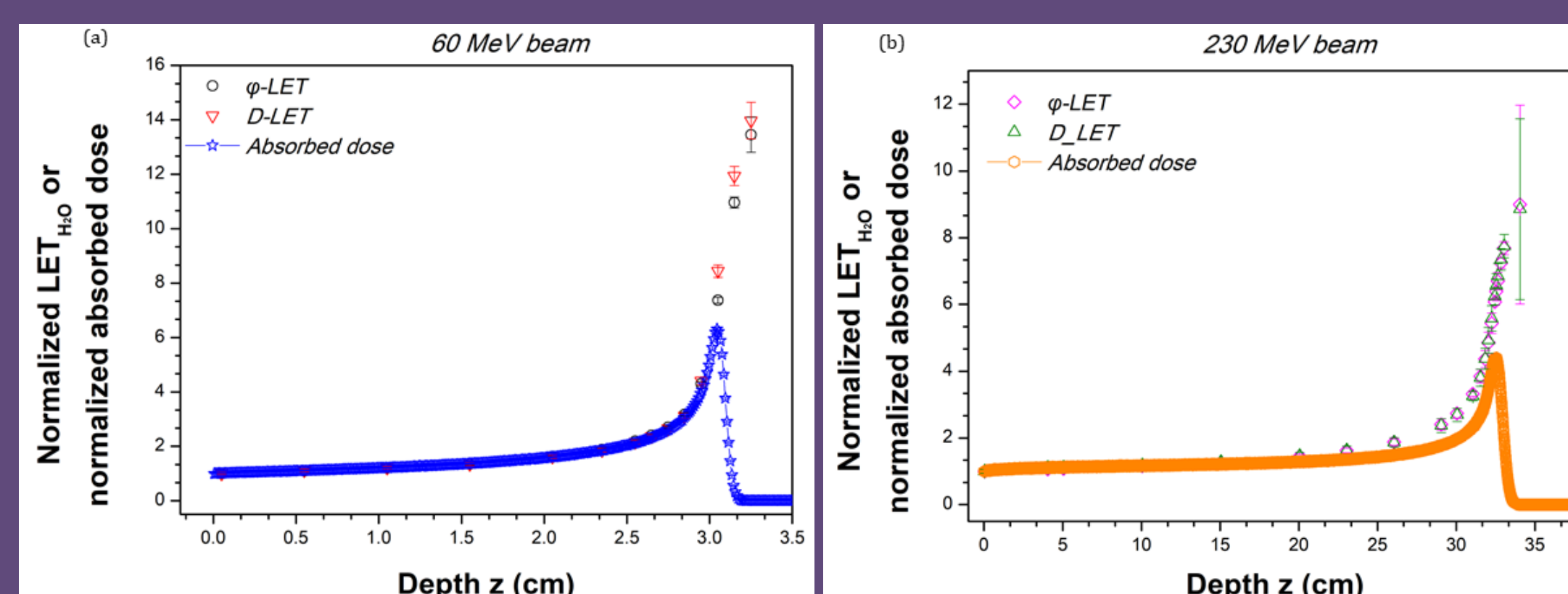


Figure 3: LETs along z-axis for 60 MeV (a) and 230 MeV (b) beam.

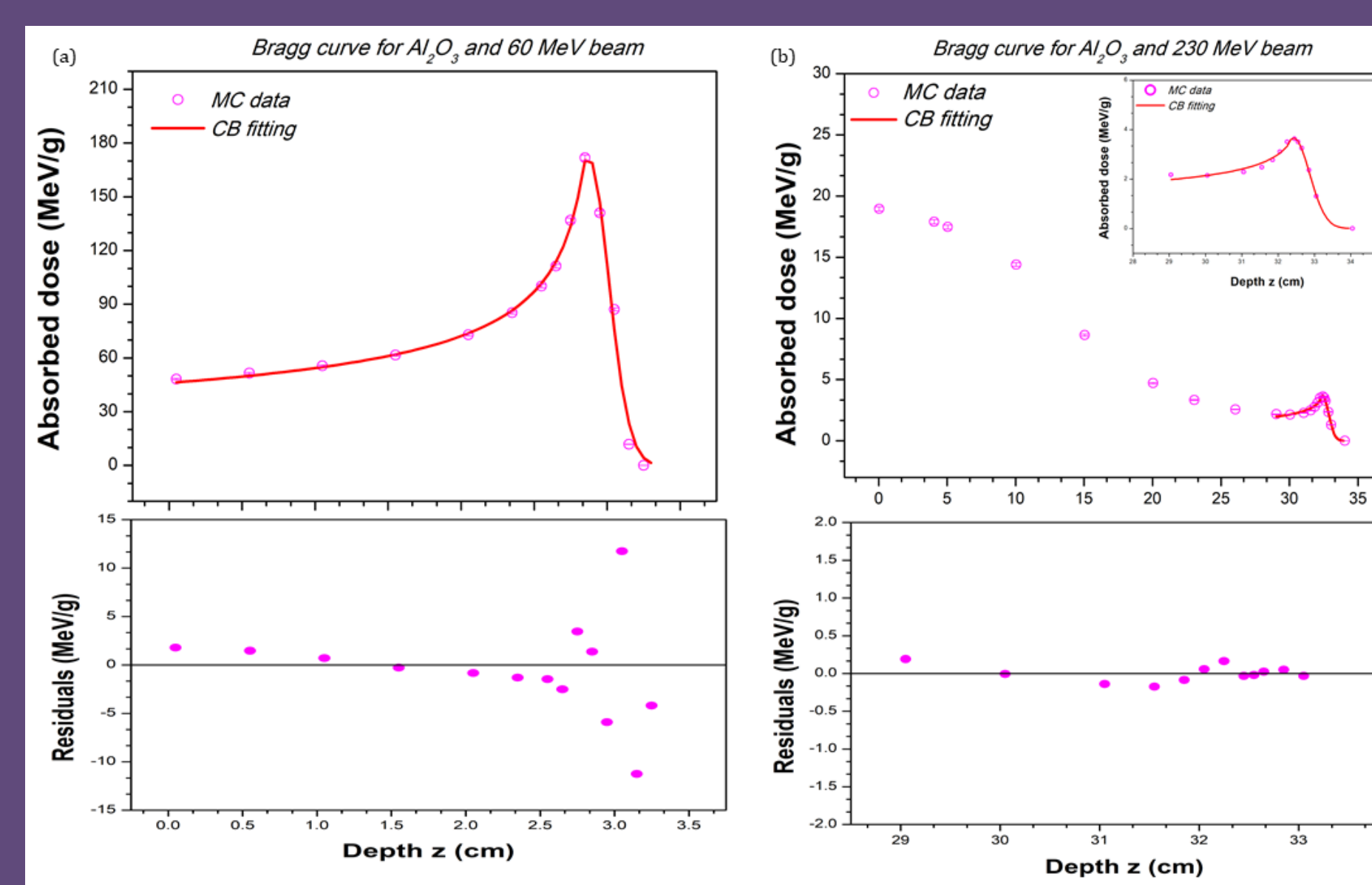


Figure 4: Fitting with CB for 60 MeV (a) and 230 MeV (b) beam, for Al₂O₃.

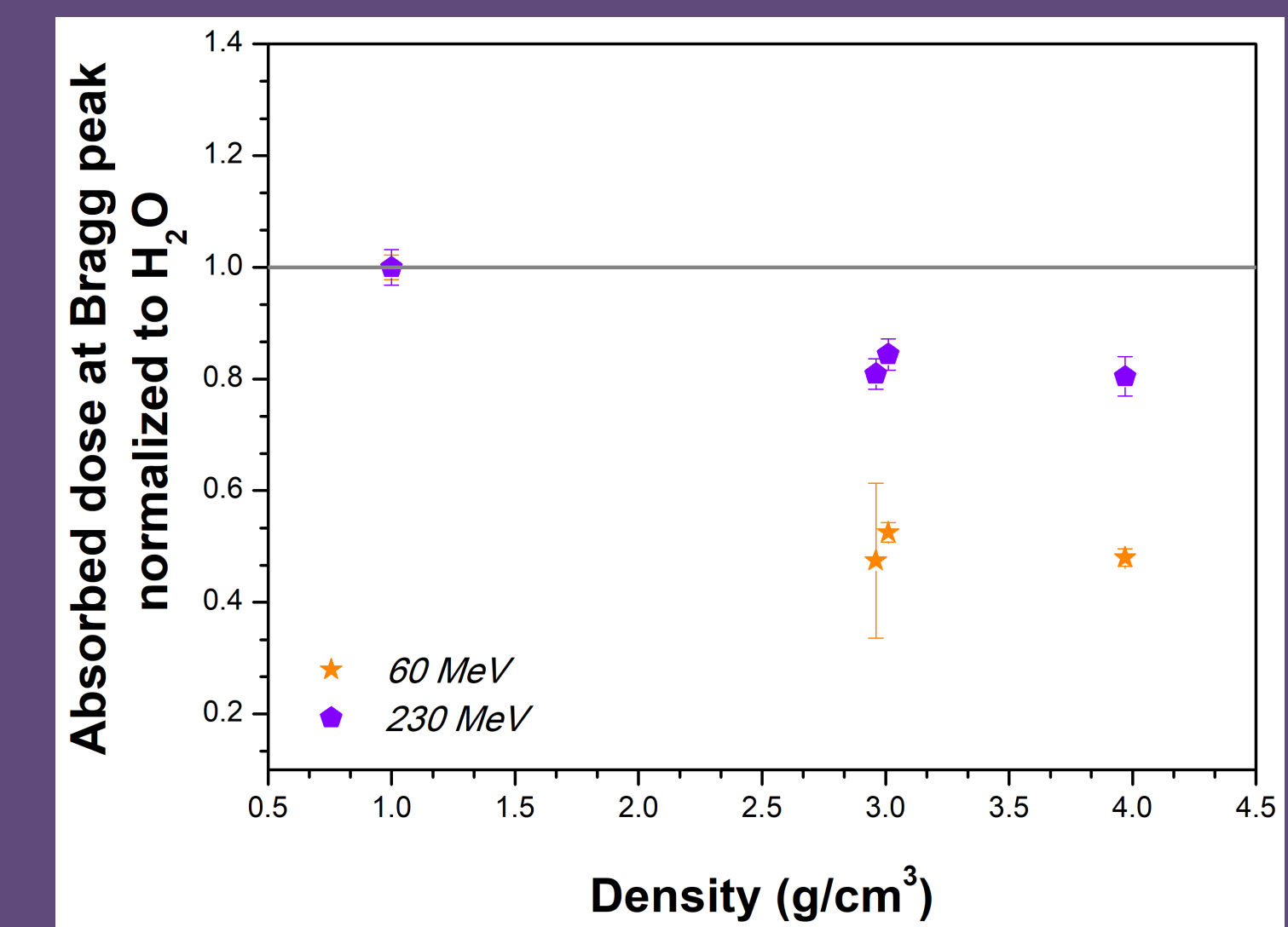


Figure 6: Absorbed dose at the peak with respect to the dosimeter's density.

Conclusions

Larger initial energy leads to larger lateral spreading, deeper Bragg peak's location and more significant energy spreading.

Proton fluence almost a Gaussian distribution for deeper locations, while there are a few low-energy protons after the peak's position.

The LETs increase rapidly at the Bragg peak's region, while they keep increasing after the peak in contrast to the absorbed dose trend.

Dosimeters with higher density present lower dose close to the peak location and shallower peak's location compared to the reference H₂O, which is a tissue equivalent material.

Future: Combination of the simulated data with luminescence measurements to calibrate the dosimeters.

Evaluation of the observed trends for more therapeutic energies and more dosimeters to obtain a mathematical model.

References

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