



EFFECTS OF PROTON BEAM DEFLECTION IN PROTON COMPUTED TOMOGRAPHY IMAGE QUALITY

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Abstract:

The treatment of tumors with radiation is a powerful tool in the fight against cancer. The radiotherapy and heavy charged particles, particularly proton-therapy, offers advantages over the more conventional methods (X-rays and gamma rays). However, all radiation treatment needs to be preceded by a planning based on some form of medical imaging, usually obtained by computed tomography X-ray (XCT), which introduces inaccuracies in planning treatment using protons. The proton computed tomography would locate, with greater certainty, the Bragg peak, ensuring the best treatment with this technique. Another advantage of proton is associated with the low dose in healthy tissues. The highest dose will be deposited in the detector. The reconstruction of images using protons depends on several factors, including the interaction of this particle with matter, which can be provided by nuclear inelastic interactions or multiple scattering Coulomb (MSC). The most likely path (MLP) and a spatial probability function of proton are the goal to be solved in treatment and planning. This work investigated this behavior of protons and the relationship with the image quality. Through calculations based on theories of Molière and Fermi-Eyegels it was verified the proton behavior over homogeneous and heterogeneous trajectories and determined the opening of this beam in places of interest. Considering the calculation in the region of the Bragg peak (6.7932 mm² - end of the peak), and the area of a square pixel of 1 mm², it is clear that the proton has a great chance to not be detected in the same entry position (straight line), which significantly affect the quality of the reconstructed image.

Index Terms: Proton-Computed Tomography, Protontherapy, Treatment Planning & Image Reconstruction

1. Introduction:

In both radiation methods treatments, proton therapy and radiation therapy, the planning of the treatment is determined by X-rays computed tomography (XCT) images 1, 2). This technique has some restrictions due the difference in the physical interactions processes of protons and photons with matter, which are associated with intrinsic properties of the target 3, 4, 5).

The interaction process of protons, which are charged particles, occurs principally by ionization and Coulomb excitation and for energies above 200 MeV by nuclear process 6, 7). The main macroscopic parameter allows to characterize the process of proton interaction with the target is called Stopping Power. This parameter also is related to the energy of the proton and the properties of the target, especially the electron density 6, 8, 9, 10).

The quality of medical image is intrinsically associated of spatial resolution and electronic density resolution 10, 11). The both quantities are related in proton Computed Tomography (pCT) with multiple scattering Coulomb and energy straggling respectively 11, 12, 13). Because of this kind of interaction with matter the proton

behavior in a target is not a straight line trajectory but a most-like path (MLP) that is dependence factor in image reconstruction process and in image quality 14, 15, 16, 17).

The Molière's theory for small angle multiple scattering give a probabilistic function about xy-proton position that is correlated with Normal distribution of scattering angle in target 17, 18). A general theory for charged particles, that loss energy and suffer multiple scattering in matter, take an approximation of a group of particles like a Gaussian beam, was proposed by Fermi and then Eygels 19). This work was aimed developed a numerical solution for Molière's and Fermi-Eygels theories about the effects of multiple scattering in image reconstruction quality.

2. Methodology:

Molière describe the scattering angle for the MLP 18)

$$\theta_{RMS} = \frac{\theta_H}{\sqrt{\chi_c B}} \quad (1)$$

To calculate the lateral deflection RMS of the charged particles beam in any position, considering the θ_{RMS} 18)

$$y_{RMS} = 3^{-1/2} \theta_{RMS} t \quad (2)$$

Scott (1963) do an approximation to B parameter $B = 1.153 + 2.583 \log \left(\frac{\chi_c}{\chi_a} \right)^2$.

Tabata and Ito (1976) 19) given a more precision definition

$$B = 2.600 + 2.3863 \log \left(\frac{\chi_c}{\chi_a} \right)^2 - \frac{3.234}{\log \left(\frac{\chi_c}{\chi_a} \right)^2 + 0.994} \quad (3)$$

Where χ_c is the characterist angle

$$\chi_c = 4\pi \int_0^t dt' \sum_i \frac{N_i(t') Z_i^2}{(\rho v)^2} \quad (4)$$

For a homogeneous target it becomes

$$\chi_c = 4\pi \frac{N}{A} t e^4 Z(Z+1) \frac{z^2}{(\rho v)^2} \quad (5)$$

N is the Avogadro number, A and Z are the relative atomic mass and atomic number respectively of the target, z, ρ and v are respectively, the atomic number, the initial velocity and the initial momentum of incident particle 18).

Molière describe an approximation expression, based on Thomas-Fermi model for the atomic potential, for χ_a , called screnning angle

$$\chi_a = \chi_o \left(1.13 + 3.76 \left(\frac{\alpha Z z}{\beta} \right)^2 \right) \quad (6)$$

where α is the fine structure constant (1/137,03604) and $\beta = v/c$. And $\chi_o = \lambda/a$, λ de Broglie wavelenght of incident particle and a the Fermi radius.

The characteristic angle scattering θ_H is a result of proton interaction with the matter, that depends the density and chemistry composition, while proton loss energy and velocity through the path.

$$\theta_H = \theta_M \sqrt{1 - \frac{0.7}{B}} \quad (7)$$

Where $\theta_M = \frac{1}{\sqrt{2}} \chi_c \sqrt{B}$ is Molière characteristic angle for multiple scattering Coulomb.

The theory of Fermi-Eygels considering a statistical variation for a beam with Gaussian distribution is given by

$$\bar{\theta}^2(x) = \bar{\theta}^2(0) + \int_0^x T(x') dx' \quad (8)$$

$$\bar{y}\bar{\theta}(x) = \bar{y}\bar{\theta}(0) + \bar{\theta}^2(0)x + \int_0^x (x-x')T(x') dx' \quad (9)$$

$$\bar{y}^2(x) = \bar{y}^2(0) + 2\bar{y}\bar{\theta}(0)x + \bar{\theta}^2(0)x^2 + \int_0^x (x-x')^2 T(x') dx' \quad (10)$$

Where x is the longitudinal position, y and θ are the deflections lateral position and scattering angle of the incident particle.

Kanematsu presented an analytical solution for homogeneous target¹⁹⁾

$$\theta^2 = f \frac{X_w}{S_o X_o} \text{Log} \left[\frac{R_o}{R} \right] \quad (11)$$

$$y = f \frac{X_w}{S_o X_o} \left(1 - \frac{R}{R_o} + \frac{R}{R_o} \text{Log} \left[\frac{R_o}{R} \right] \right) \frac{R_o}{S_o} \quad (12)$$

$$y^2 = f \frac{X_w}{S_o X_o} \left(\frac{1}{2} + \frac{3R^2}{2R_o^2} - \frac{2R}{R_o} - \frac{R^2}{R_o^2} \text{Log} \left[\left(\frac{R}{R_o} \right) \right] \right) \left(\frac{R_o}{S_o} \right)^2 \quad (13)$$

Where X_w is the radiation length of water (36.08), and X_o is the radiation length for another material; S_o is the related stopping power of the material to water; R_o is the maximum range of proton in the material and R the thickness of the target¹⁹⁾.

And f is the particle type dependent-factor

$$f = 10^{-3} z^{-0.16} \left(\frac{m}{m_p} \right)^{-0.92} \quad (14)$$

Where m is the particle mass and m_p the proton mass.

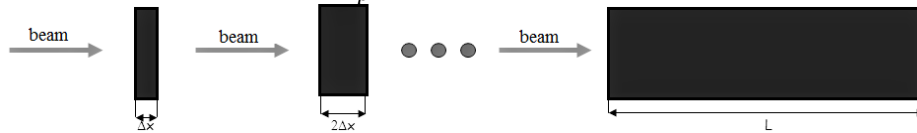


Figure 1: geometry adopted to validate the calculation. The target with water has the length of proton maximum range. Was used a pencil beam of protons.

Initially was used a target of homogeneous material (water liquid) as presented in figure 1. The target has the length of proton maximum range in water to see what happens in terms of beam scattering before and after the Bragg peak. Then was used another target of heterogeneous materials compound by air, water and bone with 200 millimeter length as shown in figure 2. Was adopted three points to observe beam scattering (35, 100, 200 mm). The calculation was done considering a proton pencil beam.

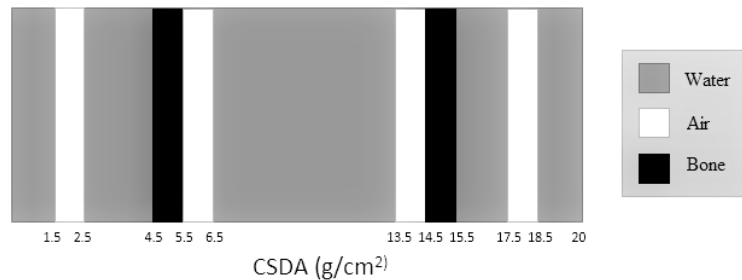


Figure 2: phantom model with water, air and bone intercalated. Was used only water to validate calculation and then this heterogeneous target.

3. Results and Discussion:

All Calculation was done for a proton beam with 200 MeV of incident energy. The maximum range with this energy for water is 25.96 g/cm², 29.37 g/cm² and 27.91 g/cm² for air and bone respectively. Table 1 shows these material compounds.

Table 1: effective atomic numbers and density of the materials used in the heterogeneous target

Material	Water (liquid)	Air	Compact bone (ICRU)
Z	10	7.6	13.8
ρ (g/cm³)	1.000	0.001225	1.850

Figure 3 shows the lateral deflection of proton beam before Bragg peak (~213 mm in water) with Molière and Fermi-Eygels calculation and after their (~259,5 mm in water) only with Fermi-Eygels calculation. Molière theory has a problem with energies below

~8 MeV, is a limitation of this model. In the other hand Fermi-Eygels is not energy-dependence but only the maximum range with the initial incident energy. Fermi-Eygels also is not mass atomic and atomic number factor-dependence like Molière, but depends on material density, most important in calculation.

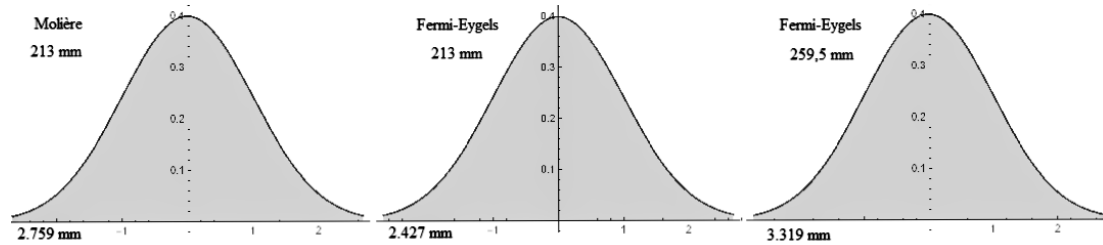


Figure 3: Lateral deflection of proton beam before Bragg peak (~213 mm in water) with Molière and Fermi-Eygels calculation and after their (~259,5 mm in water) only with Fermi-Eygels calculation.

The proton beam increases your spatial probability related with approaches the maximum range, relevant region in pCT reconstruction.

Figure 4 (a) shows the spatial probability for a proton beam inside water at 35 mm, 100 mm and 200 mm positions, calculated with Fermi-Eygels theory. And figure 4 (b) compare these values with Molière's theory. Regions faraway Bragg peak the proton beam behaves approximately like a straight line, with less deviation, for Fermi-Eygels.

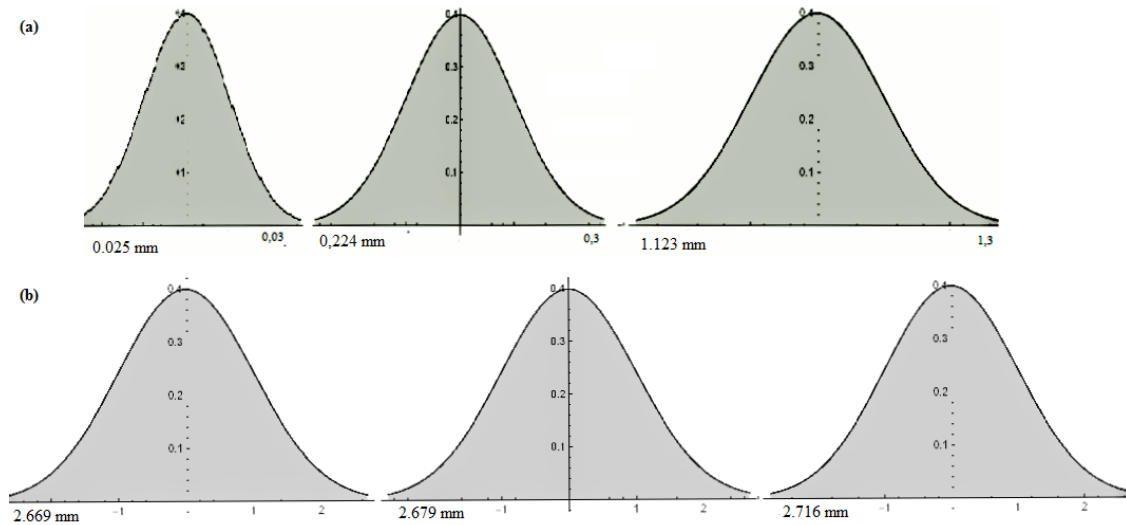


Figure 4 (a): spatial probability for a proton beam inside water at 35 mm, 100 mm and 200 mm positions, calculated with Fermi-Eygels theory; (b): spatial probability with Molière's theory at same position.

The behaviour of the lateral deflection for a pencil beam increases while proton goes inside the target. That is very important and cannot be neglectable in image reconstruction 20). For a geometry describe in figure 2 can be also analysed lateral deflection when heterogeneous target is available.

Hanson (1951) described the spatial deflection probability, considering x-y axis and the scattering characteristic angle θ_H 17).

$$\psi(x, y, z, E) = \left(\frac{1}{\sqrt{2\pi} z \theta_H(z)} \right)^2 \exp \left[-\frac{x^2 + y^2}{2z^2 \theta_H(z)^2} \right] \quad (15)$$

The calculations for the heterogeneous target give some results (to 35 mm, 100 mm and 200mm) that allow compare with the homogenous target. They are shown in figure 5.

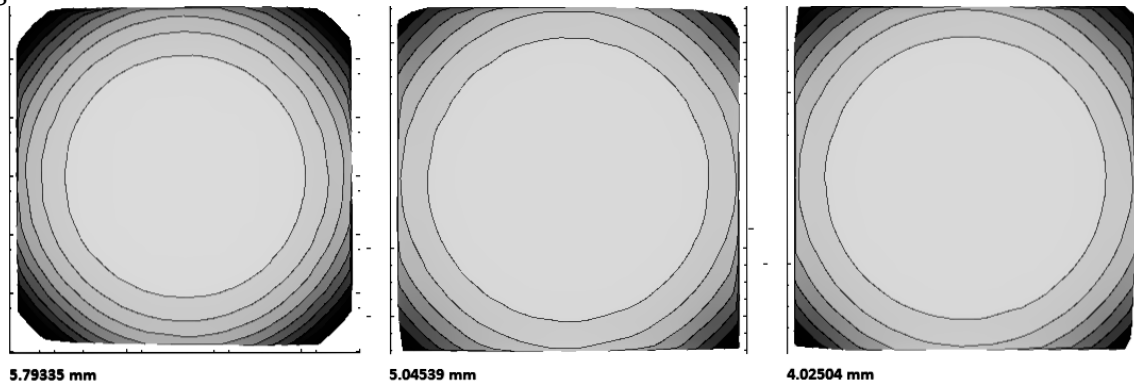


Figure 5: Spatial deflection probability at 35 mm, 100 mm and 200 mm for a target with water, air and bone.

The lateral deflection decreases in these positions and differs with the water results. It could be explained because when proton pass inside air it not undergo energy and multiple scattering what be approximate to a straight line, the interaction is drastically decreased. These results interfere directly in the lateral deflection.

4. Conclusion:

Due dose deposition proton profile the main advantage in imaging, it is to shift the maximum deposition out of the patient or upon calorimeter of the detector. However, because interactions with matter, detection of proton becomes complex.

As described, the proton has a probability space along the path, which has implications in their detection, and factor-dependence of the detector pixel size. The smaller the pixel greater is the chance of the proton detected in that position (z), have not left the same position (z) input.

Considering the calculation of spatial probability function in the Bragg peak region (6.7932 mm - the end of the peak), and the size of a square pixel of 1.00 mm² is to see that the proton has a great chance of not being detected in the same entry position.

5. References:

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