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EVALUATION OF QUALITY FACTOR FOR CLINICAL PROTON BEAMS*

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Abstract. Based on the formalism of IAEA TRS 398, the quality factor of the proton beam is a factor specific to the ionizing chamber which is correcting the difference between the quality of the reference beam and the quality of the proton beam used for therapy. The values for such a factor may be obtained by direct measurements in standard conditions or may be calculated or taken from tables, directly or by interpolations. The paper presents calculated quality factors for the situations when the experimental data are not possible or there are difficulties in directly measuring the factor for actual clinical beams using IAEA TRS 398 Code of Practice.

Key words: radiation therapy, proton therapy, quality factor (k_{QQ_0}).

1. INTRODUCTION

Radiation therapy practice has demonstrated their good therapeutic results in cancer treatment. Therefore the cancer tumors have different densities, location and radiation sensitivity. These varieties involve the necessity of using the proper radiation beam as x-rays, cobalt gamma radiation, high energy photon, electron and heavy ions beams. Because all treatment high energy beams have reference to cobalt gamma beam, these beams should be corrected to the difference between their radiation beam qualities to reference conditions. The radiation beam qualities are calculated and/or measured for proper dosimetry application in high energy photon beam [1], electron beams [2]. This paper presents the calculated values for radiation beam qualities for proton beam used in radiation therapy. The calculus has been performed according TRS 398 Code of Practice [3] for the ionization chambers owned by SSDL STARDOOR laboratory.

Proton radiation therapy is a high-precision procedure of cancer therapy, and may be considered as the most promising improvement since the introduction of

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photon and electron radiation therapy. There has been a steadily increasing interest in proton therapy over the past 10 years. Proton beams can achieve highly localized dose distributions, which should result in higher probabilities for local tumor control and disease-free survival and lower probabilities for normal tissue damage for many different tumor sites. In radiation therapy treatment is very important the technical quality. Therefore, to fully harness the power of proton radiation therapy, accurate and precise methods of dose calculation, proton range prediction, and verification of the patient position at the time of treatment are mandatory [4]. The energies considered for the incident beams are 50-250 MeV for the proton beam. These energies are reached in real treatments, by the bombarding hadrons beams, in a region located before the peak region of the Bragg curve [5].

The proton has a mass of $1.67 \cdot 10^{-27}$ kg, the electrical charge is positive and the half-life is 10^{35} years. The advantage of proton beam over the photon comes from the both of the high energy and low energy interactions. For the protons with a higher energy, the several processes of energy transfer such as direct inelastic collisions by proton, inelastic collisions by delta rays, and elastic and non-elastic nuclear reactions may occur. The secondary particles are also important, because they can be scattered to a considerable range. By these interactions, it shows characteristic depth dose curve of low dose at the entry region and high dose at a specific depth. Unlike photons or neutrons, it has a short build up region followed by a maximum energy deposition region near the end (the Bragg peak) as shown in Figure 1 [6]. As they move through target material, they interact with atomic electrons and nuclei.

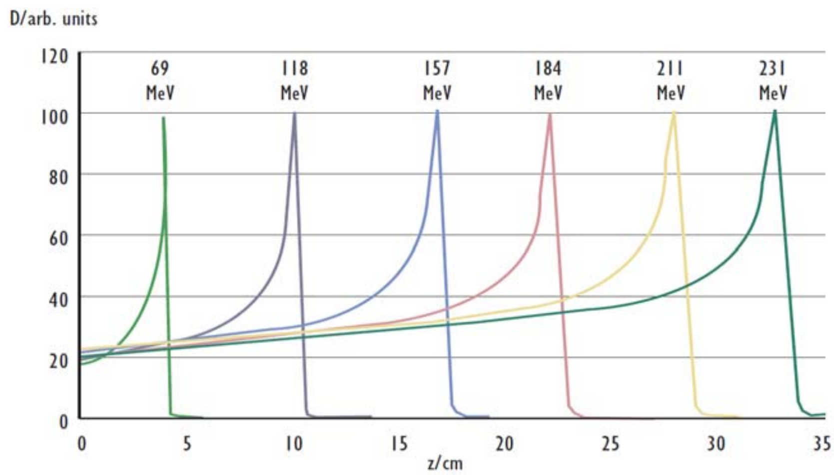

Fig. 1 – Depth in tissue curves as a function of energy for *unmodulated* proton beams [6].

The electronic interactions are ionization and excitation of atomic electrons whereas the nuclear reaction interactions are Coulomb scattering, elastic collision and non-elastic nuclear collision.

2. PROTON RADIATION BEAM QUALITY FACTOR, k_{QQ_0}, CALCULATION

Protons have a finite range and that range is energy dependent. A typical depth dose distribution for a therapeutic proton beam is shown in Fig. 2a. This consists of a region where the dose increases slowly with depth, called the 'plateau', and a region where the dose rises rapidly to a maximum, called the 'Bragg peak'. Clinical applications require a relatively uniform dose to be delivered to the volume to be treated, and for this purpose the proton beam has to be spread out both laterally and in depth. This is obtained at a treatment depth by the superposition of Bragg peaks of different intensities and energies. The technique called 'beam modulation' creates a region of high dose uniformity referred to 'spread-out Bragg peak' (SOBP) such is presented in Fig. 2b. The width of the SOBP is normally defined by the width of the 95 % dose levels. Spreading out of a Bragg peak can be achieved by different modulation techniques such as energy modulation. By appropriate selection of distribution of proton energies, a depth dose curve can be flat at ≈ 100 % over the depth of interest and then the dose falls precipitously at the end of the range to virtually zero for *modulated* proton beam. The dose decreases at the end of range from 90 % to 10 % over a distance of ≈ 0.6 mm, depending upon the range of the beam in tissue [3, 7].

The beam quality correction factor k_{QQ_0} is defined as the ratio, at the qualities Q and Q_o, of the calibration factors in terms of absorbed dose to water of the ionization chamber. The quality Q refers to high energy proton beams, and Q_o refers to cobalt gamma radiation.

Ideally, the values for k_{QQ_0} should be obtained by direct measurement of the absorbed dose at the qualities Q and Q_0, see Eq. (1), each measured under reference conditions for the user's ionization chamber used for proton dosimetry. However, at present no primary standard of absorbed dose to water for proton beams is available. Thus all values for k_{QQ_0} for proton beams are derived by calculation according Code of Practice [3] and are based on ^{60}Co gamma radiation as the reference beam quality, Q_o.,

$$k_{Q,Q_0} = \frac{N_{D,w,Q}}{N_{D,w,Q_0}} = \frac{D_{w,Q} / M_Q}{D_{w,Q_0} / M_{Q_0}} . \qquad (1)$$

The notation k_Q denotes this exclusive use of ^{60}Co as the reference quality. Values for k_{QQo} are calculated using Eq. (2)

$$k_{Q,Q_0} = \frac{(s_{w,air})_Q}{(s_{w,air})_{Q_0}} \frac{(W_{air})_Q}{(W_{air})_{Q_0}} \frac{p_Q}{p_{Q_0}} . \qquad (2)$$

W_{air} is the mean energy expended in air per ion pair formed, more usually expressed in the form W_{air}/e. The value for $(W_{\mathrm{air}}/e)_{Qo}$ in ^{60}Co, for dry air, is taken to be 33.97 J/C.

$(s_{w,\ air})_Q$ represents the Spencer-Attix water/air stopping-power ratios for beam quality Q. Q_0 refers to ^{60}Co beam. The values used are derived from the proton beam quality specifier R_{res}

$$s_{w,air} = a + bR_{res} + \frac{c}{R_{res}}, \tag{3}$$

where $a = 1.137$; $b = -4.3\cdot10^{-5}$ and $c = 1.84\cdot10^{-3}$ [3].

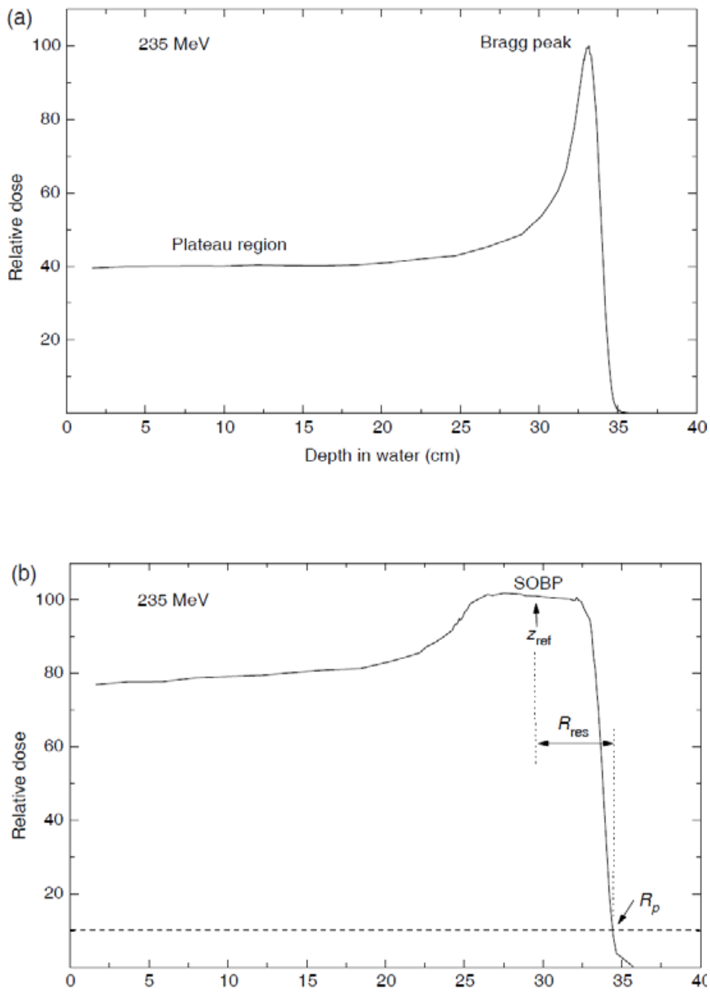

Fig. 2 – Percentage depth dose distributions for a modulated proton beam. There are indicated the reference depth z_{ref} (middle of the SOBP), the residual range at z_{ref} used to specify the quality of the beam, R_{res} and the practical range R_p [3].

The range of the beam is defined as the mean depth of penetration measured along a straight line parallel to the original direction of motion of the protons from the point where they enter the medium to the point at which additional displacement is no longer detectable. If a depth dose curve is considered, the range corresponds to the distance between the entrance surface of the beam and the distal point of 80 % dose. For clinical applications the range is usually expressed in g/cm^2, in order to be independent of specific cases where non-homogeneities can be present. If non-homogeneities are presented on the beam path (for example bone or lung), the range is slightly different from the value relative to water [3, 7].

Due to difficulties in beam energy measurement, the TRS-398 Code of Practice recommends quality specification related to residual range R_{res} [g/cm^2]. R_{res} is related to the most probable energy of the highest proton energy peak in the spectrum [8].

The residual range is measured at depth z and is defined as

$$R_{res} = R_p - z \, , \tag{4}$$

where z is the depth of measurement and R_p is the practical range (both expressed in g/cm^2), which is defined as the depth at which the absorbed dose beyond the Bragg peak or SOBP falls to 10 % of its maximum value. In the case of protons the quality Q is not unique to a particular beam, but is also determined by the reference depth z_{ref} chosen for measurement.

The uncertainty of this value was estimated by Niatel *et al.* to be 0. 2% [9]. For proton beam the value for $(W_{\text{air}}/e)_Q$ = 34.23 J/C and a standard uncertainty of 0.4 % is recommended for proton dosimetry, and these values are used in IAEA TRS 398 Code of Practice [3].

For proton beam the p_Q is considered $p_Q = 1$, with 0.4 % uncertainty.

3. RESULTS OBTAINED FOR RADIATION QUALITY, k_{QQ_0} VALUES

When a dosimeter is used in a beam of quality Q different from that used in its calibration, Q_o, the absorbed dose to water is given by

$$D_{w,Q} = M_Q N_{D,w,Q_0} k_{Q,Q_0} \, , \tag{5}$$

where the factor k_{QQ_0} corrects for the effects of the difference between the reference beam quality Q_o and the actual user quality Q, and the dosimeter reading M_Q has been corrected to the reference values of influence quantities, other than beam quality, for which the calibration factor is valid.

The calculated k_{QQ_0} correction factor values for clinical residual ranges are presented in Table 1. The k_{QQ_0} values were calculated according the Eq. (2) using the manufactured chamber data, as well as tabulated values given in IAEA TRS 398 Code of Practice. Figure 3 shows calculated values for k_Q as a function of the beam quality index R_{res} for cylindrical and plane-parallel ionization chamber types proper for proton beam dosimetry measurements, owned by STARDOOR laboratory. Calculated values for k_{QQ_0} as a function of R_{res} for other cylindrical and plane-parallel ionization chambers are given in TRS 398 [3].

Table 1

The calculated k_{QQ_0} correction factor values for clinical residual ranges

R_{res}, cm	**TN 31010**	**TN 34001**	**TN 34045**
0.25		1.008	1.009
0.5	1.032	1.004	1.005
1	1.030	1.003	1.004
1.5	1.030	1.002	1.003
2	1.029	1.002	1.003
2.5	1.029	1.002	1.003
3	1.029	1.002	1.003
3.5	1.029	1.002	1.003
4	1.029	1.001	1.002
4.5	1.029	1.001	1.002
5	1.029	1.001	1.002
7.5	1.029	1.001	1.002
10	1.028	1.001	1.002
15	1.028	1.001	1.002
20	1.028	1.000	1.001
30	1.027	1.000	1.001

Values for k_{QQ_0} for non-tabulated qualities may be obtained by interpolation between tabulated values or from Fig. 3.

The calculated values are in good agreement with tabulated values given in IAEA TRS 398 Code of Practice.

For proton beam is more suitable using a plane-parallel chamber due to its lower radiation beam quality dependence. This dependence is lower than 1.01 for both plan-parallel chambers.

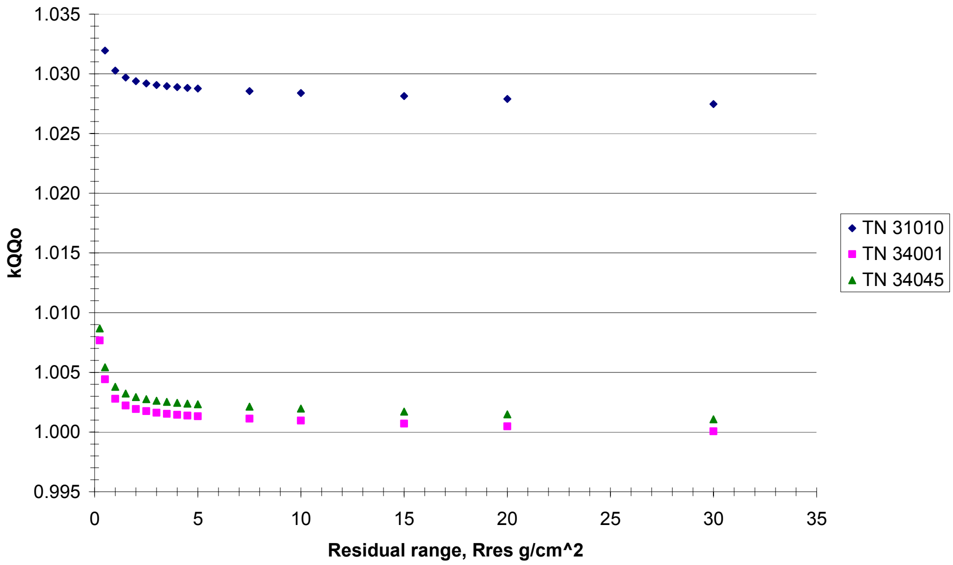

Fig. 3 – Calculated k_{QQ_0} for SSDL STARDOOR ionisation chambers types: PTW Farmer TN31010; PTW Advanced Markus TN34045 and PTW Roos TN34001.

4. CONCLUSIONS

Clinical dosimetry requires high level accuracy in dosimetry delivery system. Proton beam range is characterized by straight end point. If the radiation beam quality is not taken into consideration then the absorbed dose to target could be incorrect delivered with huge consequences for treatment results. Nowadays almost radiation therapy ionization chambers has k_{QQ_0} provided by manufacturer. In this paper we have presented the calculated k_{QQ_0} for ionization chamber proper for proton beam dosimetry owned by STARDOOR laboratory. The detailed calculus algorithm is described in TRS 398. Also TRS 398 Code of Practice provides tabulated k_{QQ_0} factors for most used chambers [3]. When tabulated data are not suitable, or not proper for particular case, then k_{QQ_0} factor could be calculated as we have specified in this paper.

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