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Assessment of radiotherapy photon beams: A practical and low cost methodology

C.Q.M. Reis*, P. Nicolucci

Departamento de Física, Faculdade de Filosofia Ciências e letras de Ribeirão Preto, Universidade de São Paulo, Av. Bandeirantes 3900, 14040-901 Ribeiro Preto, SP, Brazil

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ABSTRACT

Dosimetric properties of radiation beams used in radiotherapy are directly related to the energy spectrum produced by the treatment unit. Therefore, the development of methodologies to evaluate in a simple and accurate way the spectra of clinical beams can help establishing the quality control of the treatment. The purpose of this study is to present a practical and low cost methodology for determining primary spectra of radiotherapy photon beams from transmission measurements in attenuators of aluminum and using the method of the inverse Laplace transform. Monte Carlo simulation with PENELOPE code was used in order to evaluate and validate the reconstructed spectra by the calculation of dosimetric parameters that characterize the beam. Percentage depth dose values simulated with a 6 MV reconstructed spectrum shows maximum difference of 4.4% when compared to values measured at the corresponding clinical beam. For a 10 MV beam that difference was around 4.2%. Results obtained in this study confirm the adequacy of the proposed methodology for assessing primary photon beams produced by clinical accelerators.

1. Introduction

Ensuring the accurate delivery of prescribed dose distributions to the patient is one of the most important roles developed by clinical medical physicists. In this sense, an agreement between prescribed and delivered doses must be achieved in order to guarantee the quality of the treatment. In radiotherapy, the characteristics of depth dose distributions are directly related to the energy spectra of the beams produced by the treatment unit. Therefore, accurate calculations of doses performed by Treatment Planning Systems (TPS's) depend on detailed knowledge of energy fluence of radiation beams used in radiotherapy.

Although direct measurements of a radiation beam using a spectrometer is ideally the most accurate way to know its energy fluence (Tominaga, 1982), X-rays emitted by medical linear accelerators present high intensity and high dose rates which makes its direct evaluation technically difficult (Huang et al., 1981; Baker et al., 1995; Nisbet et al., 1998). In addition, its implementation in a clinical environment is extremely complicated and costly, since detectors with high energy resolution are relatively more expensive than ionization chambers (Baird, 1981; Delgado, 1999; Mainardi and Bonzi, 2008). In this sense, alternative methods of indirect measurement of spectra from attenuation curves of radiation on materials of known composition have been investigated over the years in order to estimate the fluence of clinical beams (Archer and Wagner, 1982; Huang et al., 1982, 1983; Boone, 1990; Baker and Peck, 1997; Waggener et al., 1999; Shimozato et al., 2007; Ali et al., 2012; Ali and Rogers, 2012). Several studies show that it is possible to relate the transmission of radiation on a material with known thickness and composition with the fluence of the original beam by making use of the inverse Laplace transform (Baird, 1981; Huang et al., 1981; Archer et al., 1985a, 1985b; Hinson and Bourland, 2002).

Obtaining the inverse Laplace transform analytically using techniques of complex analysis is not an easy process in most applications to real problems (Zakian, 1969; Brianzi and Frontini, 1991; Lien et al., 2008; Campos and Mejia Diaz, 2009). Several numerical methods have been refined over the years (Piessens, 1969; Zakian, 1970; Crump, 1976; Abate and Whitt, 1999; Rossberg, 2008; Masol and Teugels, 2010). However, comparative studies have shown that there is not an ideal method for all purposes, but methods that best fit a given problem and special circumstance (Davies and Martin, 1979; Ang et al., 1989). Therefore, the choice of a particular method should take into account, among other factors, their applicability to real inverse problems, the types of functions, the numerical accuracy desired, the computational efficiency and ease of implementation.

This work aims to present a practical and low cost methodology to

* Corresponding author.
E-mail address: fisicomelo@yahoo.com.br (C.Q.M. Reis).

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calculate the spectra of primary photon beams produced by clinical linear accelerators from attenuation measurements. The spectral reconstruction from the attenuation curve by the method of the inverse Laplace transform is evaluated and results are compared to data measured in clinical photon beams.

2. Background theory

2.1. Derivation of spectra by the inverse laplace transform

According to Hinson and Bourland (2002) when a detector is irradiated by a radiation beam, the photon fluence Φ produces a signal S that can be written as:

$$S = \int_0^\infty \Phi_E R(E) dE \tag{1}$$

where $\Phi_E = d\Phi/dE$ is the spectrum of photon fluence and the quantity $R(E)$ corresponds to the signal produced in the detector by unit fluence of photon energy E. When an attenuator of thickness x is placed between the radiation source and the detector, the signal produced changes to:

$$S(x) = \int_0^\infty \Phi_E(0) e^{-\mu(E)x} R(E) dE \tag{2}$$

where $\mu(E)$ is the attenuation coefficient of the attenuator material for photons with energy E. The relative transmission $T(x)$, defined as the ratio of the signal $S(x)$ measured in a photon beam with an attenuator of thickness x and the signal $S(0)$ measured in the same photon beam in the absence of the attenuator, can be written as:

$$T(x) = \frac{S(x)}{S(0)} = \int_0^{E_{max}} F(E) e^{-\mu(E)x} dE \tag{3}$$

where E_{max} corresponds to the maximum energy of the photon beam and $F(E)$ defined as:

$$F(E) = \frac{\Phi_E(0)R(E)}{S(0)} \tag{4}$$

is a quantity that represents the fraction of the total signal in the detector that is due to photons of energies between E and $E+dE$. A quantity $P(\mu)$ called prespectrum (Hinson and Bourland, 2002) and defined as:

$$P(\mu) = -F(E)\frac{dE}{d\mu} \tag{5}$$

can be introduced in order to change the variable of integration to $\mu(E)$.

Finally Eq. (3) can be written as:

$$T(x) = \int_0^\infty P(\mu) e^{-\mu(E)x} d\mu \tag{6}$$

which corresponds exactly to the Laplace transform of the function $P(\mu)$, so that $T(x)$ and $P(\mu)$ form a pair of Laplace transform. Since the function $T(x)$ can be determined experimentally, the calculation of its inverse Laplace transform provides the prespectrum $P(\mu)$. As a result, the spectral distribution of the beam, $F(E)$, can be determined from Eq. (5).

2.2. Solving the inverse Laplace transform

The algorithm known as the method of Stehfest (1970) uses the following expression for the approximation of the inverse Laplace transform $f(t)$ of a function $F(s)$:

$$f(t) = \frac{ln2}{t} \sum_{i=1}^{N} V_i F\left(\frac{i.\ ln2}{t}\right) \tag{7}$$

where

$$V_i = (-1)^{(N/2)+i} \sum_{k=(i+1)/2}^{min(i,N/2)} \frac{k^{N/2}(2\,K)!}{(N/2-k)!k!(k-1)!(i-k)!(2k-i)!} \tag{8}$$

and N represents a reconstruction parameter of the program, which by definition should be integer and even. Studies of numerical methods for inversion of the Laplace transform show that the Gaver-Stehfest algorithm presents good accuracy for a relatively wide range of functions, in particular when $F(s)$ is known only for real values of s or when its determination is very difficult for complex values of this variable (Davies and Martin, 1979). Theoretically, the approximate value of $f(t)$ becomes more accurate as N increases. However, in practice, rounding errors worsen the results as N becomes very large. Therefore, when using the method to evaluate an unknown function, $f(t)$, from its Laplace transform, $F(s)$, an optimal value of N must be searched in order to evaluate the accuracy of the method (Kumar, 2000). The advantage of using this algorithm is that its implementation is relatively easy, and there is no need to evaluate the function $F(s)$ for complex values of s.

3. Material and methods

3.1. Monte carlo simulations

In this study, Monte Carlo simulation with PENELOPE code was mainly used in order to optimize experimental procedures and to validate reconstructed spectra by calculating appropriate dosimetric parameters. Radiation beams were simulated in PENELOPE as point sources emitting tabulated spectra corresponding to 6 MV and 10 MV photon beams of Varian Clinac linear accelerator previously published in the literature (Sheikh-Bagheri and Rogers, 2002). Default vales of simulation parameters were used in order to obtain a reasonable compromise between speed and accuracy in all calculations (Salvat et al., 2008). In this sense, the average angular deflection parameter, C_1, and the maximum average fractional loss energy parameter, C_2, between two consecutive elastic events were set to C_1=C_2=0.1. Threshold energies for hard inelastic interactions and hard bremsstrahlung emission were set to W_{CC}=500 keV and W_{CR}=50 keV respectively. Absorption energies were set to $E_{abs}(e^-, e^+)$=500 keV for electrons and positrons and $E_{abs}(\gamma)$=50 keV for photons.

For calculating %dd curves, absorbed doses to water were evaluated along the central axis of a 60×60×60 cm^3 water phantom and using voxels of 0.5 cm thickness and 1×1 cm^2 area. A 10×10 cm^2 field was defined at a source-surface-distance (SSD) of 100 cm for both energy beams. All simulations were run with N=2×10^8 primary histories.

3.1.1. Simulation of transmission curves and spectra reconstruction

Relative transmission curves, $T(x)$, were calculated using PENELOPE code in order to evaluate the methodology of spectra reconstruction given by Eqs. (5) to (8). For this purpose, photon beams of 6 MV and 10 MV were considered to reach the surface of aluminum attenuators with a 3×3 cm^2 field at an SSD of 100 cm as illustrated in Fig. 1. Simulations were run for N_0=1×10^8 incident primary histories and a subroutine was implemented in the code in order to count the number N of primary histories that pass through the material without interacting. Values of relative transmission defined as $T(x) = N(x)/N(0)$, were calculated for thicknesses, x, of the attenuator material equals to 0.5 cm, 1.0 cm, 5.0 cm, 10 cm, 15 cm, 20 cm, 25 cm, 30 cm, 35 cm and 40 cm. The results were fitted by a non-linear function given by Archer and Wagner (1982):

$$T(x) = \left[\frac{a.\ b}{(x+a)(x+b)}\right]^{\nu}.\ e^{-\mu_m^0.x} \tag{9}$$

where a, b and ν are fit parameters. The linear attenuation coefficient of the material for the maximum energy photon component of the spectrum, μ_m^0, can also be considered as a fourth parameter to be fitted.