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Algorithmical Applications in the XiO Treatment Plan Based on Dosimetric Measurements

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Abstract

The beam modeling of the linear accelerator is performed based on dosimetric measurements by including the data obtained for the absorbed dose in water at a depth of 10 cm (D_{10}), percent depth doses, and geometric symmetry of the fields. The results obtained for photon beams with the energy 6 MV and 18 MV are placed in a database of the treatment planning system (XiO). Then, it becomes a reference system for modeling the photonic beams with high energy and creating patient treatment plans. The field modeling and calculation of the dose that should be given to the patient are done by precise dosimetric measurements. (**International Journal of Biomedicine. 2024;15(1):162-166.**)

Keywords: radiotherapy • treatment planning system • percentual depth dose • phantom

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Abbreviations

CBCT, cone-beam computed tomography; **FFT,** Fast Fourier Transformer; **MU**, monitoring units; **MGS**, multigrid superposition; **PBC**, pencil beam convolution.

Introduction

The determination of the absorbed dose for patients is based on the measurements of the dose in water because the average electronic density of soft tissues is close to the electronic density of water. Each photon beam produced by an accelerator during dosimetric measurement procedures in a water phantom is characterized by a three-dimensional dose distribution. The practical presentation of this distribution consists of three dosimetric units: in the depth dose curves, in the dose profile, and in the output factor.[1]

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The results obtained from the dosimetric measurements are then applied to create the photonic or electronic beam model, which will be applied to therapy. Also, to create this bundle model, it is necessary to know exactly what electronic density we have during the transport of the image from the scanner to the treatment plan, where the treatment bundle modeling will be performed. For this purpose, it is necessary to perform the scan of the corresponding CIRS phantom and apply the dosimetric data obtained to create the model of the treatment beam.

Methodology

This article outlines how to measure and process doses and how the XiO treatment plan works for radiotherapy. Based on the collected dosimetric data and their processing, it has

been possible to create a reference photon beam, based on which the corresponding algorithms of the therapy plan are put into operation to calculate the different doses that will be applied in patients who will undergo treatment with photon or electron beams.

The most traditional route of dose calculation in radiotherapy is through several multipurpose correction factors that describe one-by-one changes in dose and are related to changes in individual treatment parameters, such as field size and depth of dose penetration under conditions of reference. Some factors can be calculated through simple models, such as by the distance square law, to verify distance between the source of the treatment and the patient or phantom.[2]

An effective method for calculating the dose in computer calculations is the one that combines the multi-source model for finding the energy flow with that of its deposition in the patient through «convolution/superposition» algorithms. This route utilizes the natural separation between the radiation source(s) within the device head and the patient or phantom (Figure 1).

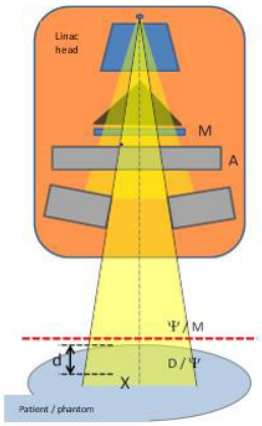

Fig. 1. *Schematic representation of the cluster formation produced by an accelerator.*

The dose calculation according to this model is given by the following equation 1:

$$\frac{D(x,d;A)}{M(A)} = \frac{\Psi(x;A)}{M(A)} * \frac{D(x,d;A)}{\Psi(x;A)} \quad (1) \ ,$$

where D is the dose, x is an arbitrary calculation point, d is the treatment depth, A represents the linac head data, M is the monitor signal, and Ψ is the power flux.This model type has the advantage of being characterized by a limited number of herd data.

The determination of the absorbed dose for clinical applications is based on the measurements of the dose in water because the average electronic density of soft tissues is close to the electronic density of water.[3] Each photon beam produced by an accelerator during dosimetric measurement procedures in a water phantom is characterized by a three-dimensional dose distribution. The practical presentation of this distribution consists of three dosimetric units: in the depth dose curves, in the dose profile, and as well as in the output factor. Measurements are usually performed with a suitable detector in tissue-equivalent phantoms, and the dose or dose rate is determined at a reference point for reference conditions.[4]

There are two types of geometries for positioning the dosimetry detector (more precisely, positioning the dosimetry detector) or the patient versus the source:

-constant geometry SSD (Source Surface Distance)

-isocentric geometry SAD (Source Axis Distance).

Because it is simpler to measure the dose with the SSD geometry, we have used exactly this geometry (Figure 2). In these geometric conditions, we place the detector, the ionization chamber, to measure the dose.

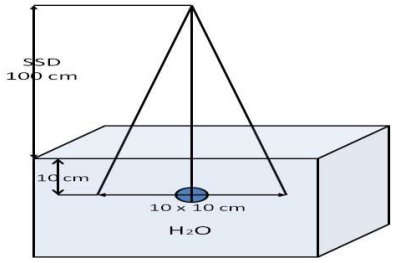

Fig. 2. *SSD geometry of dose measurement in water.*

The the absorbed dose, under standard conditions, is measured at the reference depth z_{ref} = 10cm. It is precisely this depth that is resolved since this is also where the electronic equilibrium is established, and pollution from secondary or primary electrons is almost zero. In these conditions, we have a "clean" photon bunch. The equation for measuring the dose in water is given as follows[5,6]:

$$D_{w,Q} = M * k_{T,p} * k_S * k_{pol} * k_{elec} * N_{w,Q} \ (2) \ ,$$

where M (C) is the reading of the electrometer, $k_{T,p}$ correction factor of air temperature and pressure (T_0=20$_o$C, P_0=101.3 kPa); k_S correction factor for the recombination of ions produced before they have been collected at the central electrode of the ionization chamber; k_{pol} correction factor for voltage polarity and kelec correction factor for electrometer sensitivity. $N_{(w,Q)}$ (Gy/C) is the calibration coefficient of the ionization chamber for the absorbed dose in water.[7,8] The results obtained from the dosimetric measurements are then applied to create the model of the photon or electron beam, which will be applied to therapy. Also, to create this bundle model, it is necessary to know more precisely what electronic density we have during the image transport from the scanner device to the treatment plan, in which the treatment bundle will be modeled. For this purpose, it is necessary to scan the relevant phantom with different electron densities, «CBCT Electron Density» (CIRS) (Figure 3)

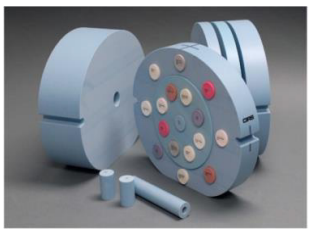

Fig.3. *Presentation of the CIRS phantom with materials of different densities.*

The obtained imaging data are placed in a database together with the dosimetric data to create the treatment herd model.

A radiotherapy treatment is administered in terms of «monitoring units» (MU), measured by two monitoring ionization chambers, independent from one another, located in the head of the linear accelerator. The purpose of dose calculation is to estimate the dose within the patient and calculate the number of MUs needed to deliver the planned dose for each treatment batch. Thus, the MU is calculated based on the following equation[3,9,10]:

$$MU = \frac{D}{CF * (PDD * OF * WF)} \quad (3)$$

D - dose (cGy)
OF – output factor (for field size 10 x 10 cm)
WF - wadge factor
CF – linac calibration factor
PDD – percentual depth dose

$$PDD(\%) = D_{max} * \left[\frac{f + d_m}{f + d}\right]^2 * e^{-\mu d} * k_{sc} \quad (4)$$

$\left[\frac{f+ d_m}{f+d}\right]^2$ expresses the law of the square of the distance

$e^{-\mu d}$ express attenuation of the photon beam

k_{sc} gives the radiation scattering factor.

The dose calculation in the XiO treatment plan is based on several algorithms, such as i) the Monte Carlo Algorithm, ii) the Fast Fourier Transformer (FFT) Convolution/MGS Algorithms, and iii) the Clarkson Algorithm.

For clinical purposes, doses are calculated as a function of time in a few minutes, and analytical models of dose calculation are used approximately. The most used analytical method for calculating the dose of photon beams is «pencil beam convolution» (PBC), Figure 4, which is given by equation 5 and «collapsed cone convolution» (CCC), Figure 5. The PBC algorithm expresses the electronic dispersion in the form of a «pencil» and dose calculations at different points of interest in the subject.

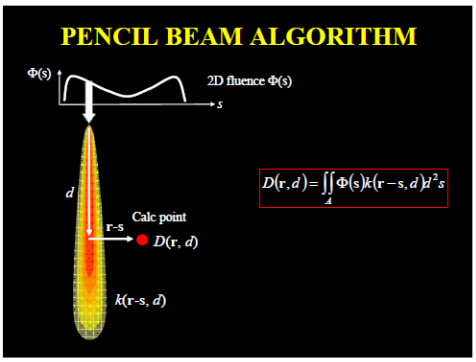

Fig. 4. *Schematic presentation of the "Pencil beam" algorithm.*

$$D(x,y,z) = \int_{-\infty}^{+\infty}\int_{-\infty}^{+\infty} S_{air}(x'',y'',z) * \frac{1}{2\pi * fmcs * \sigma_{MCS}^2(x'',y'',z)}$$
$$* exp\left\{-\frac{(x-x'')^2+(y-y'')^2}{2\pi * fmcs * \sigma_{MCS}^2(x'',y'',z)}\right\}$$
$$* G_{H2O}\left(0,0,Z_{ref}(x'',y'',z)\right)\left\{\frac{SSD_{beam} + Z_{eff}(x'',y''z)}{SSD_{beam}+z}\right\}^2 dx''dy''$$
$$+D_{photon}(x,y,Z_{eff}(x,y,z)) \quad (5)$$

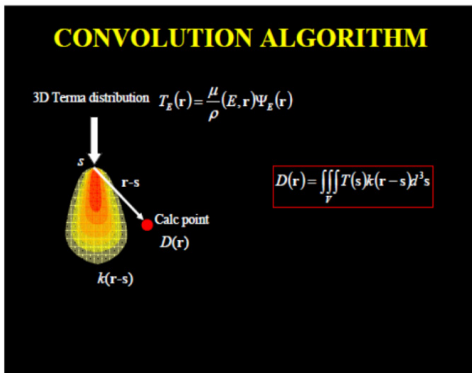

Fig. 5. *Schematic presentation of the "Convolution" algorithm.*

This paper used the "Convolution/Superposition" reference method to calculate the dose. Data obtained from direct measurements of an absorbed dose in water are configured based on the "Convolution/Superposition" algorithm to be used in radiotherapy treatment plans. The dose calculation equation with this method is:

$$D(\vec{r}) = \int T(\vec{r'}) H(\vec{r} - \vec{r'}) d^3r'$$

The FFT Convolution/Superposition reference method consists of two dose- calculation methods, but they are similar and complement each other. For this purpose, they are used interspersed in the XiO treatment plan to conform the treatment bundle to the patient's treatment plan. The FFT Convolution method is faster than the MGS method, but it does not calculate the dose accurately in the presence of non-homogeneous tissues. Due to its speed, the FFT Convolution algorithmic method is useful for batch commissioning and for calculations in homogeneous materials. In the heterogeneous case, the calculation with this algorithm is less accurate than the calculations with the old Clarkson algorithm. For patients treated in areas with non-homogeneous tissue, it is advisable not to use the FFT Convolution algorithm but another MGS algorithm.[10,11]

FFT Convolution and MGS algorithms use the same physical methodology in dose calculation. The main difference is in the way the dose is deposited:

- The FFT Convolution algorithm is presented in Cartesian coordinates, which remain unchanged throughout the space in which they are applied.
- The MGS algorithm is presented in spherical

coordinates, which are allowed to change based on the local change in the electronic density of the space in which they are applied.

The FFT Convolution and MGS algorithms use fundamental physics to describe the interaction of photons and electrons and their transport in matter. The algorithms use more fundamental theoretical physics to calculate the dose distribution. It is necessary to perform calculations for a series of reference structures to calculate the following steps:

- electronic contamination as a function of field size
- radiation spread from the phantom as a function of the cluster size
- overall radiation spread as a function of cluster size
- filter correction factor.

The superposition dose deposition method adapts the «collapsed cone convolution» analytical dose calculation method.[10] Both with the FFT Convolution method and with the MGS method, all calculations are performed for the bundle coordinates.

The dose calculated in spherical coordinates of the bundle is interpolated at each point of the specific volume, which will be used for radiation.

Results

The measurements were performed to determine the dose parameters in water and their applications in the XiO treatment system. During the processing of the dosimetric measurements, it was concluded that the dose absorbed in water at depth d_{ref} = 10 cm for a photon beam with high energy 6 MeV results in a measured dose that is 0.812 cGy/MU; and for energy 18MeV, the measured dose is 0.932 cGy/MU.

Based on these measured values, we conclude that the quality index for photon beams for energy 6 MeV is k_Q=0.990, and for energy 18 MeV, it is k_Q=0.971, when for ^{60}Co, k_Q =1.000. In accordance with these values, the LINAC apparatus is calibrated, which for 100MU gives, under standard conditions, a dose of 1.009 Gy for 6 MeV and 1.004 Gy for 18 MeV. From the scanning of the CIRS phantom in the CT suite of the Hygeia hospital for different voltages applied for scans, it was concluded that the used voltage of 120 kV gave image density close to the theoretical values with a discrepancy on the order of ±0.8% (Graph 1, Table 1, Figure 6).

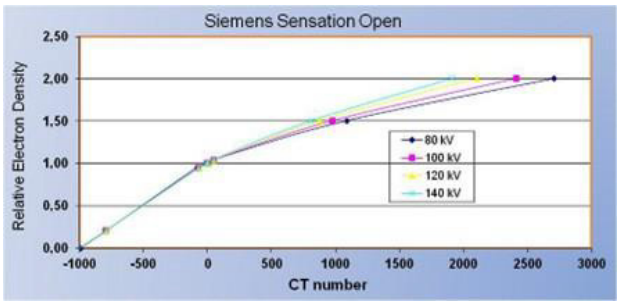

Chart 1. *Chart of densities of different materials.*

Table 1.

Different materials with measurement densities.

RED*	Material	80 kV	100 kV	120 kV	140 Kv
0.000	Air	-996	-1001	-1001	-1002
0.205	Lung	-796	-799	-798	-799
0.949	Apidose	-90	-76	-71	-68
1.000	Water	-8	-1	-2	-1
1.003	Ice	-12	-11	-12	-12
1.045	Muscle	39	44	43	42
1.506	Bone	1082	971	871	808
2.005	Bone d>	2704	2406	2102	1908
	Titanium			3069	Saturation

**RED – Relative Electron Density*

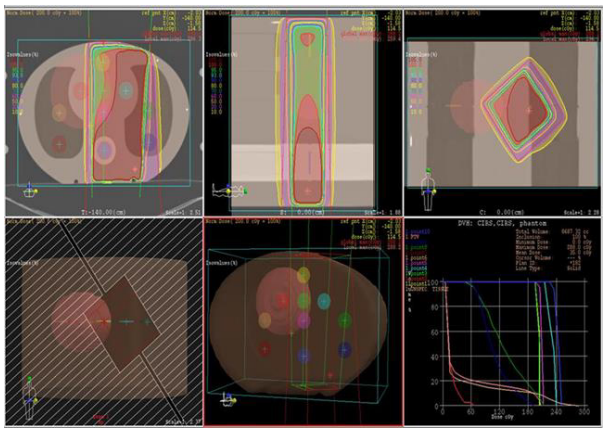

Fig. 6*. Presentation of the CIRS phantom, which is irradiated with the reference beam created based on dosimetric measurements and evaluation of electron densities of different materials.*

After receiving the dosimetric data and scanning the CBCT phantom, the reference bundle model for therapy applications is created. This bundle model is created based on the Convolution/Superposition algorithmic method.For this purpose they are used interspersed in the XiO treatment plan to conform the treatment bundle to the patient's treatment plan. The algorithmic Convolution method is faster than the MGS method, but it does not calculate the dose accurately in the presence of non-homogeneous tissues. Due to its speed, the Convolution algorithm method is useful for batch commissioning and calculations on homogeneous subjects. For patients who will be treated in non-homogeneous tissue areas, it is advisable to use the other MGS algorithm.

After processing the acquired data and placing them in a XiO treatment plan database, it was possible to create a reference set for the radiotherapy center. Based on the initial data and their processing, we concluded that in our conditions, the Convolution/Superposition algorithm best meets the requirements set in the commissioning of the linear devices.

Competing Interests

The authors declare that they have no competing interests.

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