CALIBRATION OF X-RAYS TUBES IN PRACTICAL PEAK VOLTAGE USING COMPUTATIONAL MODELING

Cassiano, D. H.[1], deise@ird.gov.br
Correa, S. C. A.[2, 3], scorrea@con.ufrj.br
Souza, E. M.[2], emsmonteiro@gmail.com
Silva, A. X. [2], ademir@con.ufrj.br
Peixoto, J. G. P. [1], guilherm@ird.gov.br
Lopes, R. T. [2], ricardo@lin.ufrj.br

[1]*Instituto de Radioproteção e Dosimetria (IRD), Comissão Nacional de Energia Nuclear (CNEN), Avenida Salvador Allende s/n Recreio dos Bandeirantes, PO Box 37750, Rio de Janeiro, 22780-160, RJ, Brasil*
[2]*Programa de Engenharia Nuclear (PEN) / COPPE, Universidade Federal do Rio de Janeiro (UFRJ), Ilha do Fundão, PO Box 68509, Rio de Janeiro, 21945-970, RJ, Brasil*
[3]*Centro Universitário Estadual da Zona Oeste, Universidade Estadual da Zona Oeste (UEZO), Avenida Manuel Caldeira de Alvarenga, 1203, Rio de Janeiro, 23070-200, RJ, Brasil*

Abstract. *The Practical Peak Voltage – PPV is adopted to measure the voltage applied to X-ray tube. The PPV is recommended by IEC 61676 (2002) and accepted and published by the code of practice TRS 457 (IAEA, 2007). The PPV is defined and applied to all forms of waves and it is related to the spectral distribution of X-rays and the properties of the image.The calibration of X-rays tubes was performed by the Monte Carlo code MCNPX. An X-ray tube for dental radiology (single phase) and X-ray tube reference (constant potential) were used in simulations for the energy range of interest: 40 kV to 100 kV and obtained results indicated a linear relationship between the tubes involved.*

Keywords*: X-Ray, PPV, MCNP, Dental Radiology, Odontology*

1. Introduction

A new quantity, known PPV – Practical Peak Voltage, was proposed by Kramer *et al*. (1998) for measure the voltage applied to X-ray tubes used on diagnostic radiology in general (IEC, 2002). The PPV was defined for a single setup applied to general radiology and after generalized to other applications of the radiodiagnostic. However, the PPV was not check up for each modality of the radiology. Believing that for applications directed to dental radiology, some considerations should be necessary to determine the PPV due to own particularity of radiodiagnostic in dental field.

The value of the PPV, for a given X-ray equipment, corresponds to the voltage applied to an X-ray equipment constant potential, which present the same image contrast obtained after the passage of X-ray beam through an acrylic phantom of 10 cm thick, with and without the addition of an aluminum attenuator filter (Kramer *et al.*, 1998; Baorong *et al.*, 2000; Ramírez-Jiménez *et al.*, 2004). At definition of the PPV, obtained images with the PPV same on different X-ray equipment that present types different of the rectification can be compared (Becker *et al*., 2003; Cassiano *et al.*, 2008). The definition of the PPV brings the values of the voltage used X-ray equipment for medical diagnosis a direct relationship between the voltage applied and the obtained contrast from the radiographic image. Obtaining the radiographic image depends on the spectra of X-ray. The spectra are dependent on several factors, especially of the high voltage applied to X-ray tube and of the beam filtration.

The dental radiology is a useful tool for diagnosis and supplements in health care (Dezotti, 2003; Van der Stelt, 1997; EC, 2004). Currently, the diagnostic radiology is considered the main source of artificial radiation in which the population is exposed (Gilboy and Fung, 2001) with approximately 14% of the total dose received each year from all sources of radiation (UNSCEAR, 2000).

In recent years a growing interest has been seen in simulating X-rays images in a number of applications (Gray, 1988; Xu *et al.*, 1999; Souza *et al.*, 2008; Correa *et al.*, 2010). The main reason for this interest is the ability to study how changes in relevant operating parameters may affect the quality of the image without the restrictions imposed by the experimental procedure and without causing exposure to radiation.

The Monte Carlo method is one of the methods employed for simulating X-ray images due to its to the stochastic nature of radiation emission, transportation and detection process. Among the main general-purpose codes based on the Monte Carlo method, the MCNPX (Monte Carlo N-Particle eXtended) code has specific functions to simulate X-ray images (Pelowitz, 2005). However, though the MCNPX code provides specific commands to simulate X-ray images, a number of procedures still must to be performed to make the MCNPX compatible with images obtained by experimental detectors (Correa *et al.*, 2010).

Because of properties mentioned, in this paper we are not making simulations of the X-ray images, but only we used the specific command to simulate X-ray image. The objective of the paper is to study a possible relationship between X-ray tube specific to dental field and X-ray tube constant potential, which is considerate as reference tube for evaluate the PPV.

2. Methods

2. 1. MC Simulations and Source Parameters

The MCNPX code employs a set of point detectors close enough to one another to create an image based on the fluence of particles passing through each detector (Pelowitz, 2005).

Usually, the MCNPX code calculates dosimetric quantities of interest by conversion factors. Thus, the MCNPX code can calculate the air kerma by the F5 command and conversion factors from fluence to kerma (DE/DF command). The DE/DF command is used to correlate obtained data in the simulation with other quantities of the interest, such as for example fluence conversion factors for air kerma through the introduction of a multiplier dependent on energy. The term DF is equal to a response function of a specific material for the incident radiation's DE energies, in MeV, in which each simulation result (fluence) is multiplied by a value of the DF conversion factor equal to the incident radiation's DE energy. Energy values outside of the range defined in the DE/DF command are interpolated between any of the higher or lower energy values contained in the respective sequences.

The distribution X-ray energy used as input parameters for simulation of the radiation beams were got through the software TASMIP (Tungsten Anode Spectral Model using Interpolating Polynomials) with tungsten anode and voltage applied to the tube varing from 40 kV to 100 kV. The spectra were provided in ranges of 10 kV. The filtration on the beam was considered a 1 mm Be (Boone and Seibert, 1997; Meyer *et al.*, 2004; Ay *et al.*, 2005; Fewell and Shuping, 1977; Boone and Chavez, 1996).

The geometry of the irradiation is shown in figure 1. A specific input file for MCNP code was developed to calculate the air kerma on the imaging detector. For values of air kerma, a punctual detector (F5 command) was positioned aligned with the central axis of the radiation beam at 50 cm distance of the acrylic phantom. The conversion factors DE/DF, provided by the ICRP 51 (1987), were used to convert fluence to air kerma. The attenuation filter (1 mm Al) was used to cover the area of the phantom before of the detector (Cassiano, 2010).

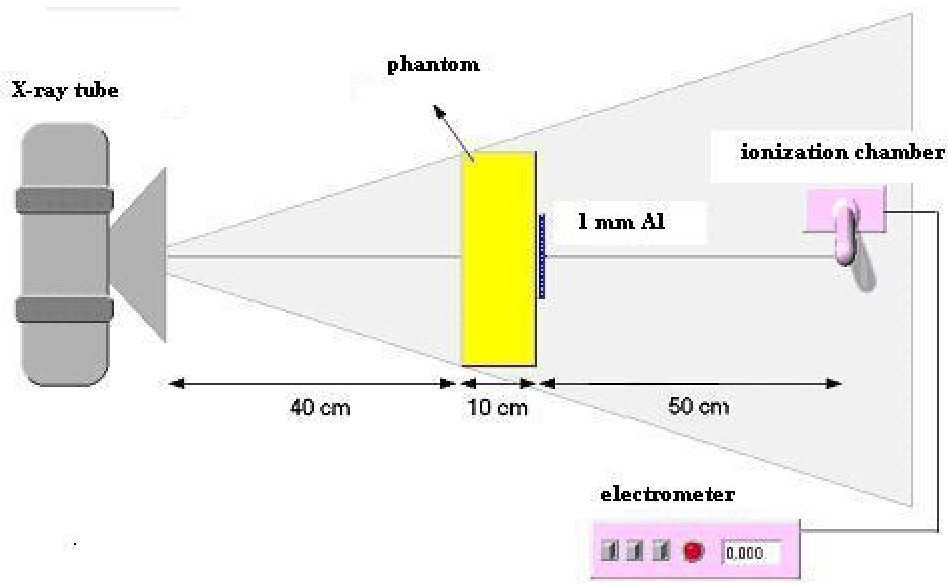

Figure 1. The diagram of the adapted setup for LNMRI´s room. It is used in simulation by the MCNPX code. The attenuator material is a plate of the aluminum with high purity (99.999%) of certification by the manufacturer.

2. 2. Calibrating

The range of voltage applied to X-ray tube for dental radiology varies from 40 kV to 110 kV and minimum effective range of voltage of the 60 kV to 90 kV with tungsten anode and additional aluminum filtration (IEC 61676, 2002). In this paper, we used voltage applied to X-ray tube from 40 kV to 100 kV, due to cover all minimum effective range of voltage that the IEC 61676 document recommends.

At this stage of the computational modeling, we consider the main types of the X-ray tubes used routinely in applications of dental radiology and we observe that X-ray tubes applied to dental radiographic in Brazil are of the single phase. For this, two X-ray tubes were used on simulations: a single-phase and ripple 100%, corresponding to the typical X-ray tube dental equipment and one X-ray tube constant potential without ripple corresponding to the X-ray tube of the LNMRI´s room. The spectra of the both X-ray tubes were provided by the TASMIP.

Thus, we study each spectral distribution of the X-ray tubes to find values close of the radiographic contrast of the phantom with the attenuator material of the 1 mm Al. We call this process of calibration of the X-ray tubes in the quantity PPV. The objective this calibration is present the radiographic properties similar of the X-ray tubes in study, so we would be compared.

In this study, the data are given by means of values of the equivalent contrast. The equivalent contrast is a form practical and precise to determinate the quantity PPV. The value of the equivalent contrast is calculated by measuring the quantity the air kerma (ICRU 33, 1980) into the point of test for two configurations (figure 1): an acrylic phantom without an attenuator material and after the same phantom with an attenuator material (aluminum plate). Obtaining value of the equivalent contrast is through of the ratio between the measurements of the air kerma at the point of test after the phantom with and without the addition of the aluminum plate (equation 1).

$$C_{K_{air}} = \frac{K_{air}^{without}}{K_{air}^{with}} \tag{1}$$

where $K_{air}^{without}$ is the value of the air kerma for the measurements without the attenuator material and K_{air}^{with} represents the value of the air kerma with the addition of the aluminum plate.

3. Results

The obtained results for the two X-ray tubes modeled by the MCNPX code are shown in table 1.

Table 1: Calibration of the X-ray tubes in the quantity PPV.

Nominal Voltage (kV)	**Equivalent Contrast**		**Percentage difference (%)**
	Constant	**Single Phase**	
40	1,3916	1,3860	0,40
50	1,2859	1,2894	0,27
60	1,2309	1,2244	0,53
70	1,1967	1,1964	0,03
80	1,1713	1,1755	0,36
90	1,1530	1,1597	0,58
100	1,1393	1,1473	0,70

The data indicated in table 1 show percentage difference less than 1% for the entire range of voltage used. This means that the MCNPX code and the software TASMIP attended, satisfactorily, all the modeling proposed in this paper. These values indicate that both X-ray tubes could be compared in environments of dental practice because they had similar radiological properties in the simulations.

During the calibration of both X-ray tubes, we observed a linear relationship with each other. The relationship is given by the equation 2:

$$y = \begin{Bmatrix} x + 5, \; para U < 60kV \\ x + 10, \; para U \geq 60kV \end{Bmatrix} \quad \textbf{[kV]} \tag{2}$$

where x is the value of voltage corresponding to the X-ray tube constant potential and y is the value of voltage of the X-ray tube with single-phase.

The equation (2) suggests that during the calibration process, the adjustment values of the voltage for the X-ray tube with single phase are higher than the values of voltage of the reference tube. This is presented as a warning sign for the routine in clinics and hospitals, because, according to the relationship above, a value of nominal voltage of a constant potential implies greater voltage in the X-ray tubes of single phase (dental radiographic) or exposure unnecessary on patient and can compromise the quality of radiographic images.

4. Conclusion

The MCNPX code was used to simulate the setup to determinate the quantity PPV applied to dental radiology. The obtained results indicated that the PPV, quantity to measure the voltage applied to X-ray tube, can be employed in dental radiology as defined by Kramer et al. (1998) and adapted for LNMRI´s room and to Brazilian scenario.

In calibration of the X-ray equipments (single phase and constant potential) the data showed percentage differences less than 1% for all observables. This has an important meaning, because these data confirm that the geometry simulated in the validation of the spectra was correctly modeled and the software TASMIP was powerful in generating different types of spectra.

5. References

Ay, M. R., Sarkar, S., Shanhriari, M., Sardari, D., Zaidi, H., 2005, "Assessment of Different Computational Models for Generation of X-Ray Spectra in Diagnostic Radiology and Mammography", Med. Phys., v. 32, n. 6, pp. 1660-1675.

Baorong, Y., Kramer, H. M., Selback, J. J., Lange, B., 2000, "Experimental Determination of the Practical Peak Voltage", The British Journal of Radiology, v. 43, pp. 641-649.

Becker, P. H. B., Ludwig, J. L., Correa, S. C. A., Peres, M. A. L., 2003, "Avaliação da Influência da Variação da Corrente do Tubo na Determinação da Grandeza Potencial de Pico Prático", In: Proceedings of the III Congresso de Metrologia.

Boone, J. M. e Chavez, A. E., 1996, "Comparison of X-Ray Cross Sections for Diagnostic and Therapeutic Medical Physics", Med. Phys., v. 23, pp. 1997-2005.

Boone, J. M., Seibert, J. A., 1997, "An Accurate Method for Computer-Generating Tungsten Anode X-Ray Spectra from 30 to 140 kV", Med. Phys., v. 24, pp. 1661-1670.

Cassiano, D. H., 2010, "Metodologia Para a Padronização da Grandeza Kerma no Ar para Aplicações em Radiologia Odontológica", Tese de Doutorado, Coordenação dos Programas de Pós-Graduação de Engenharia da Universidade Federal do Rio de Janeiro – COPPE/UFRJ, Rio de Janeiro, RJ.

Cassiano, D. H., Peixoto, J. G. P., Lopes, R. T., 2008, "Considerations on the Measurement of Peak Voltage in Dental Radiology", 8th International Symposium on Mathematical and Computational Biology, BIOMAT 2008, Campos de Jordão, SP, Brasil.

Correa, S. C. A, Souza, E. M., Silva, A. X., Cassiano, D. H., Lopes, R. T., 2010, "Computed Radiography Simulation Using the Monte Carlo Code MCNPX", Applied Radiation and Isotopes, v. 68, pp. 1662-1670.

Dezotti, M. S. G., 2003, "Avaliação de Filmes Radiográficos Periapicais em Diferentes Condições de Processamento pelos Métodos Sensitométrico, Digital e Morfométrico", Tese de Doutorado, Faculdade de Odontologia de Bauru, Universidade de São Paulo – USP, Bauru, SP, Brasil.

EC – European Commission, 2004, "European Guidelines on Radiation Protection in Dental Radiology: The Safe Use of Radiographs in Dental Practice", Radiation Protection 136, Directorate-General for Energy and Transport, Directorate H – Nuclear Safety and Safeguards, Unit H.4 – Radiation Protection 2004.

Fewell, T. R. e Shuping, R. E., 1977, "Photon Energy Distribution of Some Typical Diagnostica X-Ray Beams", Med. Phys., v. 4, pp. 187-197.

Fung, K. K. L. e Gilboy, W. B., 2001, "The Effect of Beam Tube Potential Variation on Gonad Dose to Patients During Chest Radiography Investigation Using High Sensitivity LiF: Mg, Cu, P Thermoluminescent Dosemeters, Br J Radiol, 74, pp.358-67.

Gray, J., 1988, "Three Dimensional Modeling of Projection Radiography", In: Proceedings of the Review of Progress in QNDE, 7th edition, Plenum, New York.

IAEA – International Atomic Energy Agency, 2007, "Dosimetry in Diagnostic Radiology: An International Code of Practice", Technical Reports Series N. 457, International Atomic Energy Agency, Vienna.

ICRP 51, 1987, "Data for Use in Protection Against External Radiation", ICRP Publication 51, International Commission on Radiological Protection, Oxford: Pergamon Press.

ICRU 33, 1980, "International Commission on Radiation Units and Measurements, Radiations Quantities and Units", Washington D.C.

IEC – International Electrotechnical Commission, 2002, "Medical Electrical Equipment – Dosimetric Instruments Used for Non-Invasive Measurement of X-Ray Tube Voltage in Diagnostic Radiology", IEC 61676, 1st edition.

Kramer, H. M., Selbach, H. J., Iles, W. J., 1998, "The Practical Peak Voltage of Diagnostic X-Ray Generators", Br. J. Radiol., v. 71, pp.200-209.

Meyer, P., Buffard, E., Mertz, L., Kennel, C., Constantinesco, A., Siffert, P., 2004, "Evaluation of the Use of Six Diagnostic X-Ray Spectra Computer Codes", Br. J. Radiol., v. 77, pp. 224-230.

Pelowitz, D. B., 2005, Ed. MCNPXTM User´s Manual, Version 2. 5. 0", Los Alamos National Laboratory report LA-CP-05-0369, April, 2005.

Ramírez-Jiménez, F. J., López-Callejas, R., Benítez-Read, J. S., Pacheco-Sotelo, J. O., 2004, "Considerations on the Measurement of Practical Peak Voltage in Diagnostic Radiology", The British Journal of Radiology, pp. 745-750.

Souza, E. M., Correa, S. C. A., Silva, A. X., Oliveira, D. F., Cassiano, D. H., Lopes, R. T., 2008, "Modelagem e Calibração de um Sistema de Radiografia Computadorizada Utilizando o Código de Monte Carlo MCNPX", In: XXVI CONAEND&IEV, São Paulo, SP.

UNSCEAR – United Nations Scientific Committee on the Effects of Atomic Radiation, 2000, Sources and Effect of Ionizing Radiation. UNSCEAR Reports to the General Assembly of the United Nations, with annexes. New York: United Nations.

Van der Stelt, P. F., 1997, "Radiation Protection an Quality Assurance in Dental Radiography", Ver. Belg. Med. Dent., v. 51, n. 2, pp.111-122.

Xu, J., Wallingford, R. N., Jensen, T., Gray, J. N., 1999, "Recent Developments in the X-Ray Radiography Simulation Code: XRSIM", Review of Progress in Quantitative NDE, v. 18, D.O., Thompson and D. E. Chimenti, eds., Plenum Press, New York.

6. Responsibility Notice

Cassiano, D. H.*; Correa, S. C. A.; Souza, E. M.; Silva, A. X.; Peixoto, J. G. P. and Lopes, R. T.

The authors are the only responsible for the printed material included in this paper.

* Corresponding author
E-mail address: deise@ird.gov.br (Cassiano, D. H.)