Computation and parameterization of normalized glandular dose using Geant4 (Postprint)

Authors: Kadri Omrane, Ali Alnafea Mohammed, Shamma Khaled

Date: 2023-06-18T00:00:00+00:00

Abstract

The average absorbed dose in glandular tissue is the most appropriate parameter for the assessment of the radiation-induced risk during breast imaging. The aims of this work concern: (1) the investigation of the variation effect of any related update to photon cross-section data-bases on the computation of the normalized glandular dose (DgN) for mammography quality control tests and (2) the proposition of a parameterization method leading to provide DgN values function of the breast thickness (T) and the particle energy (E) instead of E alone, as normally known. We analyzed the change effect of the photon cross-section data-bases on the computation of DgN. Those coefficients, generated using the Geant4 Monte Carlo toolkit, were studied over a range of compressed breast thickness of 2–8 cm for monoenergetic (1–120 keV by 1 keV intervals) and polyenergetic (23–35 kVp by 2 kVp intervals) X-ray beams. Moreover, breast tissue composition ranging from about 0% glandular (about 100% adipose) to 100% glandular (0% adipose) was also covered. The successful parameterization of DgN look-up table function of the breast thickness and energy, will compact its analytical form without loss of accuracy. All parameterization fits resulted in r2 values of 0.999 or better.

Full Text

Preamble

Computation and Parameterization of Normalized Glandular Dose Using Geant4

Omrane Kadri,[1],[2] Mohammed Ali Alnafea,[1],† and Khaled Shamma[1]
[1]Department of Radiological Sciences, College of Applied Medical Sciences, King Saud University, PO Box 10219, Riyadh 11433, Saudi Arabia
[2]National Center for Nuclear Sciences and Technologies, 2020 Tunis, Tunisia

(Received July 18, 2014; accepted in revised form September 16, 2014; published online June 20, 2015)

The average absorbed dose in glandular tissue is the most appropriate parameter for assessing radiation-induced risk during breast imaging. This work aims to: (1) investigate how updates to photon cross-section databases affect the computation of normalized glandular dose (DgN) for mammography quality control tests, and (2) propose a parameterization method that provides DgN values as a function of both breast thickness (T) and particle energy (E), rather than E alone as is conventional. We analyzed the effect of photon cross-section database changes on DgN computation. These coefficients, generated using the Geant4 Monte Carlo toolkit, were studied across compressed breast thicknesses of 2–8 cm for monoenergetic (1–120 keV in 1 keV intervals) and polyenergetic (23–35 kVp in 2 kVp intervals) X-ray beams. Additionally, breast tissue composition ranging from approximately 0% glandular (100% adipose) to 100% glandular (0% adipose) was covered. Successful parameterization of the DgN look-up table as a function of breast thickness and energy compacted its analytical form without loss of accuracy. All parameterization fits achieved r^2 values of 0.999 or better.

Keywords: Monte Carlo simulation, Geant4, Normalized glandular dose, Parameterization
DOI: 10.13538/j.1001-8042/nst.26.030303

Introduction

Since glandular breast tissue is regarded as the most radiosensitive tissue, breast dosimetry has been considered an important basis for assessing radiation risk to patients undergoing mammography [1]. The average absorbed dose in glandular breast tissue, known as the mean glandular dose (MGD), is preferred for radiation risk assessment. It can be estimated by measuring the entrance air kerma and applying normalized glandular dose (DgN) conversion coefficients for a given breast thickness and glandularity [2].

Among the most prominent factors influencing DgN values is the photon fluence per exposure factor (photon quanta per mm^2 per mR). This factor depends on mass energy absorption coefficient tables for air (μen/) that can be extracted from existing databases such as MCPLIB [3] and XCOM [4]. Moreover, as Zoetelief et al. [2] pointed out, using MCPLIB cross sections yields DgN values approximately 10% higher than using XCOM data.

Thus, there is a need to update existing monoenergetic DgN(E) look-up tables whenever changes occur in the (μen/) tables, which immediately affect the Monte Carlo simulation programs used to generate such data. On the other hand, due to the large volume of tabulated DgN values used for optimizing mammographic procedures provided by many authors [5–10], parameterization is clearly required to simplify direct MGD calculation after measuring incident air kerma. Many works have focused on this issue [11–13]. However, introducing breast thickness as a second parameter beyond energy for database parameterization does not appear to be well studied, which could lead to more generalized fitting equations.

This paper examines the effect of the photon fluence per exposure parameter on DgN calculations—a factor that appears essential after observing variations in (μen/) values between data derived from Hubbell et al. [14] and those used by Boone et al. [11]. We parameterize monoenergetic DgN conversion coefficient data so that simple custom functions can be easily implemented in spreadsheets to calculate DgN values for given input parameters (photon energy, breast composition, and thickness). We adopt a mathematical fitting method that includes breast thickness as a second parameter and compare its accuracy in predicting direct computational results. Finally, we evaluate the method' s ability to estimate MGD for clinically relevant cases, as performed by many authors [15], using X-ray tube anode/filter combinations of Mo/Mo (30 μm), W/Rh (50 μm), and W/Pd (50 μm) at 23–35 kVp, with a breast thickness of 4 cm and glandularity of 50%. This work can be considered a continuation of recent studies on MGD calculation [16–18] performed by medical physicists as part of mammography quality control tests.

II. Materials and Methods

Many Monte Carlo simulation packages can be used to estimate MGD, each with different advantages and disadvantages. Accurate and versatile general-purpose packages such as Geant3 [19], EGS4 [20], MCNP [21], and most recently Geant4 [22] include well-validated physics models, geometry-modeling tools, and efficient visualization utilities that require no additional tailoring for medical imaging applications. For this study, we employed the Geant4 toolkit.

To provide an updated dataset of DgN coefficients for mammography covering the energy interval of 1-120 keV, thicknesses of 2–8 cm, and glandularities of 0%-100% for compressed breast, we adopted the same experimental setup used by Boone et al. [11]. The studied energy interval enables use of look-up tables for standard (up to 40 kVp) and dual-energy (up to 120 kVp) mammography imaging. We developed a Monte Carlo simulation program to generate the data, followed by a mathematical fitting procedure to parameterize the large dataset.

According to Refs. [1, 23, 24], MGD calculation is highly influenced by the spatial distribution of glandular tissues within the breast. Thus, this study is dedicated to further enhancing quality assurance and quality control procedures involving equipment performance, comparisons of X-ray machine efficiency, and mammography dosimetry protocols. For larger compressed breast thicknesses up to 8 cm, the same parameterization procedure should be followed, with the penalty of more complex equations requiring additional fitting parameters.

A. Simulation Procedure

To derive DgN(E) values, we used the photon quanta per mm^2 per mR parameter, K(E), for a given energy E (in keV) of an incident photon beam in air medium as derived by Johns et al. [25]:

$$K(E) = 54300/[E(\mu_{en}/\rho)_{air}]$$

where $(\mu_{en}/\rho)_{air}$, in cm^2/g, is the mass energy absorption coefficient in air.

The DgN(E) coefficient, in mGy/mGy, is given by [26, 27]:

$$DgN(E) = 1.8352 \times 10^{-17} E_{dep} G(E) K(E) f_g \rho (T - 2T_{skin})(R_2/R_1)^2$$

where E_{dep} is the energy deposited in breast tissue, f_g is the glandular fraction of the breast, ρ is breast density, $R_1 = 8$ cm is breast radius, $R_2 = 7.6$ cm is breast tissue radius excluding skin, T_{skin} is skin thickness, T is breast thickness, and $G(E)$ is a factor describing the proportion of absorbed dose in glandular tissue to overall breast tissue, given by:

$$G(E) = f_g(\mu_{en}/\rho)_g f_g(\mu_{en}/\rho)_g + (1 - f_g)(\mu_{en}/\rho)_a$$

where $(\mu_{en}/\rho)_g$ and $(\mu_{en}/\rho)_a$ are the mass energy absorption coefficients for glandular and adipose tissue, respectively, for the given energy. The unit of the constant 1.8352×10^{-17} is (mGy · g)/keV.

Using recent mass energy attenuation coefficients for photons in air provided by NIST [14], we tailored a Monte Carlo simulation program to calculate absorbed energy in the breast model for given configurations of breast tissue composition, thickness, and monoenergetic photon beams. An additional procedure was added to compute the $G(E)$ factor during simulation runtime, which outputs DgN(E) coefficients for each studied case.

The Geant4-based simulation program modeled monochromatic X-ray emission from a "point-like" source and tracked particle transport through the breast. Physical processes included the photoelectric effect, Compton and Rayleigh scattering for photons, and bremsstrahlung, ionization, and "multiple scattering" for electrons. Production thresholds were set to 1 keV for photons and 10 keV for electrons. From the existing suite of physical packages ready for use within the simulation program, "PhysListEmStandard" was selected. Elemental compositions of adipose, glandular, and skin tissue were based on Ref. [27].

For each combination of (monoenergetic) X-ray energy, breast size, and breast composition, a simulation run was executed using 10^6 incident photons, providing sufficient statistical precision. To simplify the analytical form of fitting functions and due to the small contribution to MGD parameters, we omitted the energy interval of 1-11 keV during the database fitting stage. This contribution does not exceed 1% of the MGD for the studied X-ray spectra. DgN(E) coefficients for 11-120 keV X-rays were computed.

A total of at least 1582 ($113 \times 7 \times 2$) runs were carried out for database generation, covering 113 energy values (11-120 keV) and seven thicknesses (2-8 cm) for glandularities of 100% and 50%. The absorbed dose by breast tissue was recorded for direct derivation of DgN(E) coefficients (in SI units of mGy/mGy).

Following the same simulation setup used by Boone et al. [11], we used a source-to-detector distance of 60 cm. Changing this distance to 80 cm or 40 cm would deviate beam energy at the breast vicinity by less than 1%, as adding or subtracting a 20 cm air layer affects the primary beam trajectory by 0.4% and 0.06% for 30 keV and 100 keV photon energies, respectively. Moreover, modeling the X-ray tube focal spot geometry as point-like rather than disc is valid to within approximately 1%. Replacing a point-like source with a disc of Φ40 µm or Φ300 µm (as found in clinically realistic tests) deviates DgN(E) by 0.213% or 0.444% for photon energy, glandularity, and thickness combinations of (50 keV, 50%, 2 cm) or (10 keV, 100%, 8 cm), respectively, though focal spot size affects image contrast [28].

We consider the homogeneous distribution of glandular tissues within the breast a limitation of the current model, as Dance et al. [29] noted that absorbed energy by glandular tissue depends on glandular tissue position within the breast.

The simulation was performed using Geant4 (V9.4.p01) on a Red Hat Enterprise Linux 5 workstation with an Intel Core i7 CPU, 4 GB RAM, running at 3.40 GHz. The statistical uncertainty (2σ) associated with all Monte Carlo calculations presented in this work is less than 1%.

B. Parameterization Procedure

The following parameterization procedure was adopted to reduce the large volume of DgN values while retaining prediction precision to an acceptable accuracy level using a simple analytical fitting equation. For accuracy assessment, the r^2 statistical coefficient was calculated. We fitted the complete DgN(T,E) dataset using Eq. (4) for the energy interval 11-120 keV, for glandularities of 100% and 50% separately:

$$DgN(E) = a + c\ln(T) + e\ln(E) + g\ln^2(T) + i\ln^2(E) + k\ln(T)\ln(E)1 + b\ln(T) + d\ln(E) + f\ln^2(T) + h\ln^2(E) + j\ln(T$$

MGD was calculated using Eq. (5):

$$\mathrm{MGD} = \frac{\sum_{E_{min}}^{E_{max}} K(E) \times DgN(E) \times E \times \phi(E)}{\sum_{E_{min}}^{E_{max}} K(E) \times \phi(E)}$$

where $\phi(E)$ is photon fluence per exposure in units of mR per photon per mm^2, and E_{min} and E_{max} are the minimum and maximum energy spectrum limits.

Physically, Eq. (4) shows that any DgN coefficient explicitly depends on incident photon energy and breast phantom thickness [30]. Thus, we preferred fitting these values with a bivariate polynomial depending on E and T [13]. The quotient form arises because the DgN coefficient represents a fraction of absorbed dose in the breast model relative to air medium, each being a polynomial energy-dependent function. To achieve relative differences between computed and fitted values not exceeding 3% in MGD calculation for most X-ray beam spectra, we adopted the second-order polynomial fitting Eq. (4). Greater precision would require extension to higher order, with the penalty of a more complex fitting expression.

To our knowledge, fitting methods using parameter E have been treated similarly by many authors [7, 26, 30], whereas introducing parameter T as a second variable in its current form is proposed here for the first time.

III. Results and Discussion

We calculated the photon fluence per exposure conversion coefficient using Eq. (1) for photon energies of 1-120 keV. Figure 1 [Figure 1: see original paper] shows the relative difference between $K(E)$ calculated based on the most recent mass energy absorption coefficient data for air provided by NIST [14] and those tabulated by Boone et al. [26]. The major differences occur for photons with energies below 75 keV, reaching 12% at 27 keV as shown in the inset. For photon energies above 75 keV, this difference does not exceed 1%.

We began by studying the effects of the photon fluence per exposure coefficient, which caused modifications in DgN values between those of Boone et al. [11] and those presented in this work. Additionally, fitting parameters were tabulated for easy handling of the computational model. Based on the parameterization procedure, we evaluated the efficiency of reproducing directly simulated MGD for specific cases. We concluded by analyzing the interpolation and extrapolation capabilities of this procedure. Future work could address heterogeneous breast tissue composition and prone breast positioning.

B. K(E) Variation Effect on Boone' s Look-up Table

Table 1 compares MGD calculations between literature (Boone et al. [5]) and the present study using updated $K(E)$ values. The comparison examines three anode/filter combinations: Mo/Mo (30 µm), W/Rh (50 µm), and W/Pd (50 µm) at 23-35 kVp with HVL of 0.269-0.597 mm Al. The breast thickness was 4 cm and glandularity was 50%. We observed maximum relative differences between literature and Geant4 data of approximately 1.9%, 5.0%, and 4.2%, and between calculated (parameterized) and Geant4 data of approximately 2.6%, 3.8%, and 4.1%, respectively, for the above anode/filter combinations.

Referring to Mo and W experimental spectra in [31], we observed that the two characteristic peaks were located around 17 keV and 11 keV, respectively, as shown in Fig. 2 [Figure 2: see original paper]. The spectra covered 23–35 kVp. Figure 2 also illustrates curves of relative error (%) for $K(E)$, $DgN(E)$, and their sum. According to Eq. (5) and Fig. 2, we expect the MGD relative difference for tungsten anode spectra to be twice or higher than that for molybdenum. This can be explained by the coincidence of the tungsten peak with the maximum relative deviation. Furthermore, we can infer that the relative deviation for rhodium anode spectra will be lower than for molybdenum within the studied kVp interval.

C. Data Set Generation and Parameterization

The complete normalized glandular dose dataset was generated using Eqs. (2) and (3) and the developed simulation code for breast glandularities of 100% and 50%. The dataset corresponding to $f_g = 50\%$ is shown in Fig. 3 [Figure 3: see original paper]; plots for $f_g = 100\%$ are similar. Table 2 presents the parameters resulting from fitting the complete DgN values using Eq. (4) for the energy interval 11–120 keV. No subdivision of energy and thickness intervals was needed for parameterization purposes. Statistical test p-values of 0.9999 were obtained for both cases.

D. MGD Prediction Ability

To investigate MGD prediction capability using the current DgN dataset with the fitting method, we used Eq. (5) and photon fluence per exposure spectra corresponding to anode/filter combinations of Mo/Mo (30 µm), W/Rh (50 µm), and W/Pd (50 µm) generated according to the MASMIP and TASMIP Boone codes [26] for various kVp and HVL X-ray tube parameters. The heel effect associated with the spectra was not considered in this study, similar to [16]. Thus, the beam was assumed to have uniform intensity and spectral quality within the field covering the entire breast. If accounted for, MGD would vary as it is proportional to the fluence spectra.

From the data at 35 kVp in Table 1, we observe that MGD values become similar due to the small contribution of DgN for low photon energies at high kVp. The results are in acceptable agreement with those directly simulated.

E. Interpolation and Extrapolation Capability

The DgN data values in Table 3 demonstrate the effectiveness of interpolation and extrapolation procedures in terms of glandularity. Direct calculations using linear interpolation and extrapolation techniques and Monte Carlo simulation were conducted for various glandularity cases, with data tabulated. We find acceptable agreement between the two datasets, allowing straightforward interpolation and extrapolation of MGD for any given glandularity from approximately 0% to 100%.

IV. Conclusion

We calculated the photon fluence per exposure conversion coefficient using Eq. (1) for photon energies ranging from 1 keV to 120 keV. A computer simulation program was developed to estimate radiation dose to breast phantoms during quality control and quality assurance mammography tests. Monoenergetic DgN coefficients were computed for various energies and thicknesses, and the data were parameterized and presented as a general equation with parameters depending on two variables: E and T. The Geant4-based DgN coefficients presented in this work provide an updated version of those currently used in mammographic dosimetry quality control tests. The ultimate need to recompute such coefficients relates to any updates in mass-energy absorption databases for materials, which in turn influence all Monte Carlo simulation outputs. Although the relative difference found in $K(E)$ parameter calculation reached 12% at 27 keV, the relative difference in MGD values attained only 5% for the studied molybdenum and tungsten anode spectra. The parameterization procedure allowed us to correctly compute MGD values for any X-ray spectra. Using linear interpolation and extrapolation techniques, we can vary kVp up to 120 keV, breast phantom thickness from 2 cm to 8 cm, and glandularity proportion from approximately 0% to 100%, correctly reproducing MGD values derived from direct simulation. Such parameterization can be straightforwardly applied to modern techniques such as breast CT and digital breast tomosynthesis, though compressed breast CT is of particular concern.

References

[1] Nigapruke K, Puwanich P, Phaisangittisakul N, et al. Monte Carlo simulation of average glandular dose and an investigation of influencing factors. *J Radiat Res*, 2010, 51: 441–448. DOI: 10.1269/jrr.10008

[2] Zoetelief J and Jansen J Th M. Calculation of air kerma to average glandular tissue dose conversion factors for mammography. *Radiat Prot Dosim*, 1995, 57: 397–400.

[3] Everett C J and Cashwell E D. MCP code fluorescence-routine revision. Technical report LA-5240-MS, Los Alamos Scientific Laboratory, 1973. DOI: 10.2172/4411256

[4] Berger M J and Hubbell J H. XCOM: Photon cross-sections on a personal computer. National Bureau of Standards Gaithersburg, MD, USA, version 1.2, 1987. DOI: 10.2172/6016002

[5] Boone J M. Glandular breast dose for monoenergetic and high-energy X-ray beams: Monte Carlo assessment. *Radiology*, 1999, 213: 23–37. DOI: 10.1148/radiology.213.1.r99oc3923

[6] Thacker S C and Glick S J. Normalized glandular dose (DgN) coefficients for flat-panel CT breast imaging. *Phys Med Biol*, 2004, 49: 5433–5444. DOI: 10.1088/0031-9155/49/24/003

[7] Dance D R. The Monte Carlo calculation of integral radiation dose in xeromammography. *Phys Med Biol*, 1980, 25: 25–37. DOI: 10.1088/0031-9155/25/1/003

[8] Dance D R, Young K C and van Engen R E. Estimation of mean glandular dose for breast tomosynthesis: Factors for use with the UK, European and IAEA breast dosimetry protocols. *Phys Med Biol*, 2011, 56: 453–471. DOI: 10.1088/0031-9155/56/2/011

[9] Dance D R. Monte Carlo calculation of conversion factors for the estimation of mean glandular breast dose. *Phys Med Biol*, 1990, 35: 1211–1219. DOI: 10.1088/0031-9155/35/9/002

[10] Dance D R, Young K C and van Engen R E. Further factors for the estimation of mean glandular dose using the United Kingdom, European and IAEA breast dosimetry protocols. *Phys Med Biol*, 2009, 54: 4361–4372. DOI: 10.1088/0031-9155/54/14/002

[11] Boone J M. Normalized glandular dose (DgN) coefficients for arbitrary X-ray spectra in mammography: computer-fit values of Monte Carlo derived data. *Med Phys*, 2002, 29: 869–875. DOI: 10.1118/1.1472499

[12] Sobol W T and Wu X Z. Parametrization of mammography normalized average glandular dose tables. *Med Phys*, 1997, 24: 547–554. DOI: 10.1118/1.597937

[13] Li X, Zhang D and Liu B. A parameterization method and application in breast tomosynthesis dosimetry. *Med Phys*, 2013, 40: 092105. DOI: 10.1118/1.4818059

[14] Hubbell J H and Seltzer S M. Tables of X-ray mass attenuation coefficients and mass energy-absorption coefficients (version 1.4). National Institute of Standards and Technology, Gaithersburg, MD, 2004. http://physics.nist.gov/xaamdi

[15] Sato H, Ando M and Shimao D. Investigation of absorbed radiation dose in refraction-enhanced breast tomosynthesis by a Laue case analyser. *Radiat Prot Dosim*, 2011, 146: 231–233. DOI: 10.1093/rpd/ncr157

[16] Ma A K W, Darambara D G, Stewart A, et al. Mean glandular dose estimation using MCNPX for a digital breast tomosynthesis system with tungsten/aluminium and tungsten/aluminium + silver X-ray anode-filter combinations. *Med Phys*, 2008, 35: 5278–5289. DOI: 10.1118/1.3002310

[17] Ciraj-Bjelac O, Beciric S, Arandjic D, et al. Mammography radiation dose: initial results from Serbia based on mean glandular dose assessment for phantoms and patients. *Radiat Prot Dosim*, 2010, 140: 75–80. DOI: 10.1093/rpd/ncq040

[18] Dance D R and Young K C. Estimation of mean glandular dose for contrast enhanced digital mammography: factors for use with the UK, European

and IAEA breast dosimetry protocols. *Phys Med Biol*, 2014, 59: 2127–. DOI: 10.1088/0031-9155/59/9/2127

[19] Brun R, Bruyant F, Maire M, et al. GEANT3 user guide. CERN report DD/EE/84-1, Geneva, 1986.

[20] Nelson W R, Hirayama H and Rogers D W O. The EGS4 code system. Technical report SLAC-265, SLAC Linear Accelerator Center, 1985.

[21] Briesmeister J F. MCNP–A general Monte Carlo N-particle transport code. Technical report LA-12625-M, Version 4B, Los Alamos National Laboratory, 1997.

[22] Agostinelli S, Allison J, Amako K, et al. Geant4–a simulation toolkit. *Nucl Instrum Meth A*, 2003, 506: 250–303. DOI: 10.1016/S0168-9002(03)01368-8

[23] Koutalonis M, Delis H, Spyrou G, et al. Monte Carlo generated conversion factors for the estimation of average glandular dose in contact and magnification mammography. *Phys Med Biol*, 2006, 51: 5539–5548. DOI: 10.1088/0031-9155/51/21/010

[24] Vedantham S, Shi L X, Karellas A, et al. Dedicated breast CT: radiation dose for circle-plus-line trajectory. *Med Phys*, 2012, 39: 1530–1541. DOI: 10.1118/1.3688197

[25] Johns H E and Cunningham J R. *The Physics of Radiology*. Springfield, Illinois: Charles C Thomas, 1974.

[26] Boone J M and Anthony Seibert J. An accurate method for computer-generating tungsten anode X-ray spectra from 30 to 140 kV. *Med Phys*, 1997, 24: 1661–1670. DOI: 10.1118/1.597953

[27] Hammerstein G R, Miller D W, White D R, et al. Absorbed radiation dose in mammography. *Radiology*, 1979, 130: 485–491. DOI: 10.1148/130.2.485

[28] Law J. The influence of focal spot size on image resolution and test phantom scores in mammography. *Brit J Radiol*, 1993, 66: 441–446. DOI: 10.1259/0007-1285-66-785-441

[29] Dance D R, Hunt R A, Bakic P R, et al. Breast dosimetry using high-resolution voxel phantoms. *Radiat Prot Dosim*, 2005, 114: 359–363. DOI: 10.1093/rpd/nch510

[30] Okunade A A. Evaluation of average glandular dose in mammography. *Med Phys*, 2006, 33: 1153–1164. DOI: 10.1118/1.2179150

[31] Boone J M, Fewell T R, Jennings R J, et al. Molybdenum, rhodium, and tungsten anode spectral models using interpolating polynomials with application to mammography. *Med Phys*, 1997, 24: 1863–1874. DOI: 10.1118/1.598100

Note: Figure translations are in progress. See original paper for figures.