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EVALUATION OF EXPOSURE IN MAMMOGRAPHY: LIMITATIONS OF AVERAGE GLANDULAR DOSE AND PROPOSAL OF A NEW QUANTITY†

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The radiation risk in mammography is traditionally evaluated using the average glandular dose. This quantity for the average breast has proven to be useful for population statistics and to compare exposure techniques and systems. However it is not indicating the individual radiation risk based on the individual glandular amount and distribution. Simulations of exposures were performed for six appropriate virtual phantoms with varying glandular amount and distribution. The individualised average glandular dose (iAGD), i.e. the individual glandular absorbed energy divided by the mass of the gland, and the glandular imparted energy (GIE), i.e. the glandular absorbed energy, were computed. Both quantities were evaluated for their capability to take into account the glandular amount and distribution. As expected, the results have demonstrated that iAGD reflects only the distribution, while GIE reflects both the glandular amount and distribution. Therefore GIE is a good candidate for individual radiation risk assessment.

INTRODUCTION

As the glandular tissue is considered to be the tissue at risk in the breast, the radiation risk in mammography is usually evaluated by the mean dose to the glandular tissue for an average breast, also designated as average glandular dose (AGD). This is in agreement with the recommendations of ICRP 103 (2007): 'the mean value of absorbed dose as averaged over tissue can be correlated with radiation detriment for stochastic effects in that tissue' and to evaluate 'equivalent doses for the reference male and female'. Recommendations of quality control organisms[1] suggest to use the computation proposed by Dance *et al.*[2], consisting of a factorisation of the AGD in the incident air kerma and several conversion factors. Dance *et al.*[3], Wu *et al.*[4], Boone[5] and Sechopoulos *et al.*[6] calculated their conversion factors from Monte Carlo simulations on a semi-circular or cylindrical breast phantom. In different articles[2, 3, 7, 8] the conversion factor of Dance *et al.* was developed to take into account the various breast thicknesses (g-factor), massic densities (c-factor), anode/filter materials (s-factor) and recently the projection angles in breast tomosynthesis (t-factor). These factors were provided as a function of thickness and beam quality expressed in terms of half value layer. Wu *et al.*[4, 9] tabulated the normalised glandular dose, D_{gN}, for different breast densities and added tables for different anode–filter combinations. Hammerstein *et al.*[10] suggested that 'mean dose to gland for the "average" breast can be used as a basis for comparing doses delivered with different radiographic techniques'. This definition neither takes into account the local variations in dose due to the heterogeneous distribution of glandular tissue, in particular in the direction of the X-ray beam, nor the amount of glandular tissue if it deviates significantly from the average. However, several limitations appear when attempting to use an individualised average glandular dose (iAGD) for an individualised risk estimation as highlighted by Sechopoulos *et al.*[11]. The main limitation of the AGD was indicated by Dance *et al.*[12] who showed a difference up to 43 % between the AGD computed by simulations for a textured phantom and the AGD for the same phantom computed from the tables for homogeneous phantoms. Also Porras-Chaverri *et al.*[13] found a considerable difference in computed AGD for a phantom with a total density of 50 % but consisting of three layers of different densities. Differences between the case-specific calculated AGD and the AGD based on the published tables can go up to 50 % for phantoms of 8 cm compressed breast thickness with the densest layer on the down side. On patient cases Sechopoulos *et al.*[11] found an overestimation of 27 % on average for the AGD computed for the structured breast compared with a same amount of homogeneous density. Another limitation of individualisation of the AGD concerns the amount of glandular tissue of the average breast. The concept of a 50 % glandular breast was introduced by Hammerstein *et al.*[10] in

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1979 in a first attempt to estimate roughly the amount and distribution of an average breast. However since then it was shown that the average breast density at the age of the screening population is rather around 20 %[14–16].

The concept of the AGD as dose to the glandular tissue of the average breast is well-suited for quality control and population surveys, but is not fulfilling the requirements of an individual risk measure[17]. Therefore in their article Hammerstein *et al.*[10] proposed the 'total energy absorbed in glandular tissue (E_g) as the most relevant indicator of risk in mammography'. This is comparable to the imparted energy proposed by Huda *et al.*[18] for dosimetry in computed tomography (CT) imaging. However the main constraint at that time was also reported by Hammerstein *et al.*[10]: 'detailed information will have to be obtained on the amount and distribution of gland tissue in many individual cases before E_g can be applied properly to the problem of individual risk'. Volumetric breast density computation methods solved the problem of the amount of glandular tissue[19, 20]. Recent developments in breast imaging as breast tomosynthesis and breast CT might partly overcome the problem of the localisation of the glandular tissue. It becomes thus possible to provide an individualised quantitative evaluation of the radiation risk in mammography with an improved quantity, based on the amount and distribution of glandular tissue.

The goal of this study is to reintroduce the E_g concept of Hammerstein *et al.* under the name glandular imparted energy (GIE) as an improvement in the quantification of the individual radiation-induced risk in mammography. It was evaluated through simulations of exposures to six virtual phantoms.

MATERIALS AND METHODS

Phantoms

Simulations were performed for six semi-circular phantoms (see Figure 1). Phantom 1 is the reference breast model used by Dance[3]. It is a semi-cylinder of 4.5 cm thick with a radius of 8 cm. The central region is composed of a 50 % adipose–50 % glandular homogeneous mixture by volume, surrounded on all sides except the chest-wall side by 0.5 cm adipose tissue, representing the skin. Phantom 2 differs from phantom 1 by its radius of 5.8 cm. Taking into account the 0.5 cm adipose skin layer, the amount of the projected glandular tissue of phantom 2 is therefore divided by two compared with phantom 1. Phantom 3 is similar to the reference phantom except that the central region is exclusively composed of adipose tissue. Phantoms 4, 5 and 6 have the same size and shape as phantoms 1 and 3 (8 cm radius, 4.5 cm thickness) and contain the same amount of glandular tissue as phantom 1 (0.151 kg), but with a different distribution. All the glandular tissue is gathered into a homogeneous 1.75 cm plate, positioned at mid-breast height (phantom 4), in the upper part just below the skin layer (phantom 5) and in the lower part just above the skin layer (phantom 6).

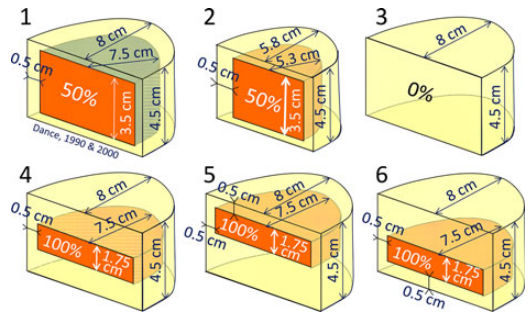

Figure 1. Six phantom configurations for which Monte Carlo simulations are performed to compute iAGD and GIE.

Quantities

AGD is used as the organ dose for collective risk evaluation based on the linear-non-threshold hypothesis (LNT). Using the method proposed by Dance *et al.*[2], the AGD is obtained by computing the product of the air kerma and the three conversion factors: one for the phantom thickness, one for the density and one for the spectrum. The phantom thickness and the spectrum are the same for all phantoms and exposures in the experiment. The density to be used in the Dance tables is the massic density in the central, intra-skin compartment. For the phantoms the intra-skin density was defined to be always 50 % glandular–50 % adipose tissue by volume, except for phantom 3, where it is 0 %. The conversion from volumetric (VG) to massic density (MG) can be done using

$$\mathrm{MG} = \frac{\rho_G}{(1-\mathrm{VG})\rho_A + \mathrm{VG}\rho_G}\mathrm{VG}, \qquad (1)$$

with $\rho_A = 0.93$ g cm^{-3} and $\rho_G = 1.04$ g cm^{-3}, the volumetric mass densities of adipose and fibroglandular tissue from Hammerstein *et al.*[10]. The resulting density to be used for the AGD computations is 53 %.

The definition of the iAGD was derived from the usual AGD and obtained by computing the imparted energy in the glandular tissue for the specific case, then dividing it by its glandular mass. Considering that the glandular tissue is distributed over the entire breast, the iAGD remains consistent with the ICRP concept of mean value of absorbed dose averaged over the tissue.

The LNT model is based on fundamental cellular processes coupled with dose–response data. This means that the radiation-induced risk for the total gland can be computed by integrating the risk over all

the individual cells of the gland. Assuming the same mass for all cells and integrating the cellular dose over all cells provide a quantity proportional to the total energy imparted to the gland (GIE). This meets the original proposal of Hammerstein describing 'total energy absorbed in glandular tissue as the most relevant indicator of risk in mammography'. As GIE is the total energy absorbed by the glandular tissue, it is expressed in joules (J). To facilitate computations it can be normalised to the incident air kerma and is then expressed in milli-joules per milli-gray (mJ mGy^{-1}).

Dose computations

The energy and doses delivered to the different phantoms were computed using the dose module of the Monte Carlo simulation platform CatDose[21]. Doses normalised to an incident air kerma of 1 mGy were computed for 28 kV molybdenum target/molybdenum filter. The AGD, iAGD and GIE normalised to the entrance air kerma were computed for all phantoms, as well as the ratios of the AGD, the iAGD and the GIE for phantoms 2–6 relative to those of phantom 1.

RESULTS

The results of the simulations are presented in Table 1. As expected, the AGD is the same for phantoms 1, 2, 4, 5 and 6. However it is slightly higher for phantom 3 in spite no tissue at risk is present, demonstrating the strong limitation of the AGD for individual cases.

iAGD

The iAGD of phantoms 1 and 2 are the same, whereas the amount of glandular tissue in phantom 2 is only half that of phantom 1. However the iAGD takes into account the different positions of the same total amount of glandular tissue along the X-ray axis with the same projected area: the iAGD for phantoms 4–6 changes by almost a factor 7 between the highest values (gland close to the entrance surface, iAGD = 0.312) and the lowest values (gland close to the exit surface, iAGD = 0.045) of the different phantoms. In this experiment the iAGD is sensitive to the position of the glandular tissue, but not to its amount.

GIE

The GIE of phantom 2 is divided by two compared with phantom 1, reflecting the ratio of the glandular contents. The GIE for phantom 3 is 0, in agreement with the absence of glandular tissue and associated risk. The GIE for phantoms 1, 4, 5 and 6 are quite different (0.24–1.66 times that of phantom 1), in spite they share the same glandular mass. The GIE thus expresses the energy effectively received by the glandular tissue, the same way as iAGD does.

DISCUSSION

The AGD has been originally introduced for comparing doses delivered with different radiographic techniques and was further extended to assess the radiation risk in mammography. The results of this study demonstrate the limitations of the AGD for individualised risk assessment since it does not take into account the individual glandular amount and distribution.

Individualising the AGD by taking into account the glandular amount and distribution allows to effectively compute the delivered energy to the gland in each point and thus to differentiate the phantoms and their risks. Extending the concept to real breasts requires the 3D-localisation of the glandular tissue over the breast, which became possible only recently with new 3D imaging techniques such as tomosynthesis and breast CT. To make it operable in daily clinical practice, methods should be developed to estimate the local energy absorption without performing a Monte Carlo simulation for each case.

The iAGD converts the delivered energy to the gland into a dose in the usual way and takes into account the glandular distribution. However its low sensitivity to the amount of tissue at risk demonstrates that it is not sufficiently effective for individual risk assessment.

Therefore GIE is a better alternative. Introducing the GIE in clinical practice might cause some discomfort compared with the current AGD, in particular due to the change in nature and units. It should be

Table 1. Results of the simulations.

Phantom number	1	2	3	4	5	6
AGD (mGy mGy^{-1})	0.188	0.188	0.237	0.188	0.188	0.188
Relative AGD[a]	1	1	1.26	1	1	1
iAGD (mGy mGy^{-1})	0.188	0.188	—	0.110	0.312	0.045
Relative[a] iAGD	1	1	—	0.59	1.66	0.24
GIE (mJ mGy^{-1})	0.0285	0.0142	0	0.0167	0.0473	0.0068
Relative[a] GIE	1	0.5	0	0.59	1.66	0.24

[a]Relative to the results of phantom 1.

noted that computing the GIE from the AGD for phantom 1 can be done by multiplying AGD and the glandular mass: 0.188 mGy mGy^{-1} × 0.151 kg = 0.0285 mJ mGy^{-1}. However for other glandular distributions the results may be different. For example, for phantom 4 the GIE is 0.0167 mJ mGy^{-1} for the same AGD and the same glandular mass as for phantom 1. Of the three evaluated quantities, the GIE satisfies the best the needs for individualised quantification of the radiation-induced risk.

CONCLUSION

The goal of this study was to propose a radiation quantity suitable to individual risk assessment[21]. The GIE depends on both the amount and distribution of the glandular tissue, confirming its expected capability to provide a quantitative evaluation of the individual radiation risk in mammography.

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