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BREAST DOSIMETRY USING HIGH-RESOLUTION VOXEL PHANTOMS

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A computer model of X-ray mammography has been developed, which uses quasi-realistic high-resolution voxel phantoms to simulate the breast. The phantoms have 400 μm voxels and simulate the three-dimensional distributions of adipose and fibroglandular tissues, Cooper's ligaments, ducts and skin and allow the estimation of dose to individual tissues. Calculations of the incident air kerma to mean glandular dose conversion factor, *g*, were made using a Mo/Mo spectrum at 28 kV for eight phantoms in the thickness range 40–80 mm and of varying glandularity. The values differed from standard tabulations used for breast dosimetry by up to 43%, because of the different spatial distribution of glandular tissue within the breast. To study this further, additional voxel phantoms were constructed, which gave variations of between 9 and 59% compared with standard values. For accurate breast dosimetry, it is therefore very important to take the distribution of glandular tissues into account.

INTRODUCTION

A Monte Carlo computer program has been developed to realistically model mammographic X-ray imaging systems. The model uses a voxelised phantom to simulate the breast and takes account of the various components of the imaging system including the X-ray spectrum, compression plate, anti-scatter device and image receptor. Anatomical details can be included in the voxel phantom and the program can be used to estimate measures of image quality[(1)]. The program also calculates the doses to the different tissues simulated by the breast voxel phantom and in particular to the glandular tissues, so that the mean glandular dose (MGD) may be estimated. The MGD is believed to be related to the risk of radiation-induced carcinogenesis, and is the quantity normally used for breast dosimetry. When the MGD is calculated in combination with measures of image quality, the model can be used for optimisation.

This paper focuses on the use of the program and a series of voxel phantoms for breast dosimetry. At present, the European protocol for breast dosimetry[(2)] is based on the calculations of Dance[(3)]. Later work by the same group[(4)] is used in the United Kingdom to facilitate calculations for breasts of varying glandularity. The computer code used for these calculations and that used in the United States for the same purpose[(5,6)] were based on very simple models of the breast. The use of such calculations will therefore result in a systematic error in the estimation of the MGD for any particular breast, or the average MGD for a population. In view of the worldwide use of mammography for both screening and the examination of symptomatic women, it is important to study the limitation of these simple models. The results presented here are based on two series of voxel phantoms, which allow changes in the distribution of glandular tissue to be made and the effect of these changes on MGD to be studied.

METHODS

Monte Carlo model

The Monte Carlo computer program is based on programs developed previously by our group[(4,7)], which have been extended by the addition of a voxel phantom. The program follows photons from the focal spot of the X-ray tube, through the compression plate and into the breast, simulating photoelelectric interactions and coherent and incoherent scattering. All energy deposited within each breast tissue is recorded so that the dose to individual tissues and the MGD can be estimated. The incident air kerma at the upper surface of the breast (without backscatter) is also calculated.

Breast dosimetry

In the European and United Kingdom Mammography protocols, the MGD, D_G, for individual patients is estimated from experimentally determined values of the incident air kerma at the upper surface of the breast, K_i, with the help of conversion coefficients estimated from Monte Carlo calculation by using a simple model of the breast (Figure 1). This simple

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model assumes that the compressed breast is a cylinder of semi-circular cross section with a central region, which is a uniform mixture of adipose and glandular tissues surrounded by an adipose shield of 5 mm thick. The MGD is given by:

$$D_G = K_i g, \quad (1)$$

where the conversion coefficient g is the ratio of the MGD for the particular breast under consideration to the incident air kerma. This coefficient depends on the breast model used and the incident X-ray spectrum. All data presented here are for a Mo/Mo X-ray spectrum at 28 kV with a half-value layer (HVL) of 0.357 mm aluminium.

Voxel phantoms

Two series of voxel phantoms and calculations have been used to study the variation of the conversion coefficient, g, with the distribution of the glandular tissue within the breast. The first series of calculations used the breast voxel phantoms developed by Bakic *et al.*[8]. Here, these phantoms are referred to as 'structured phantoms'. They simulate the uncompressed breast with three regions. The central region is connected to the nipple and contains glandular tissue, adipose tissue and a ductal tree. The second region surrounds the central region and contains adipose tissues and Cooper's ligaments. Here, it is referred to as the 'surround region'. The third region contains skin. The breast is modelled using random numbers to select the position and size of structures within each region according to chosen distribution laws. A voxel size of 400 μm was used for the present work. Eight phantoms constructed in this way were used for the first series of calculations. They corresponded to breast thicknesses in the range 40–80 mm and glandularities in the range of 25–100%. The term 'glandularity' refers to the fraction by weight of glandular and ductal tissues within the central region of the breast, rather than the fraction by weight for the whole breast. This definition is consistent with that used previously for the calculations for simple breast models, although it is not well suited to some of the situations simulated.

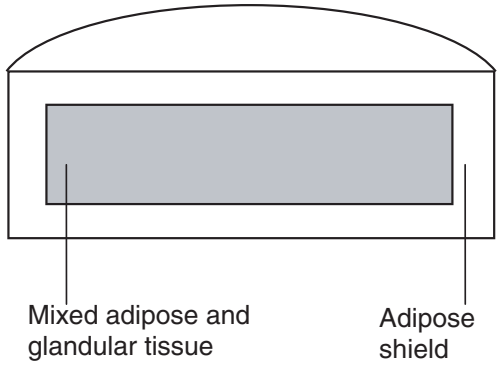

Figure 1. Simple geometrical model of the breast used previously for the calculation of the MGD.

The eight structured phantoms only enabled a limited study of the effect of the distribution of the glandular tissue on the conversion coefficient, g, and it was necessary to develop a further series of phantoms in order to study the effect of greater changes in the distribution. The new phantoms, referred to as 'unstructured phantoms' were based on those of Bakic *et al.*[8], and had central, surround and skin regions. However, the central and surround regions were filled voxel-by-voxel rather than structure-by-structure. The composition of each voxel was chosen at random from the tissues present in the region. For the surround region, the average composition used by Bakic *et al.*[8] was maintained. For the central region, the glandularity could be varied to generate different phantoms. Figure 2 shows vertical slices though a 50 mm structured phantom and the unstructured phantom which simulates it. Varying distributions of glandular tissue were obtained by moving the boundary between the central and surround regions of the phantoms, maintaining connectivity with the nipple. Two series of calculations were performed. In the first series, both the upper and lower boundaries between the central and surround regions were raised in such a way that the volume of the central region and connectivity with the nipple were maintained. In the second series, the upper and lower boundaries were moved symmetrically about the horizontal plane through the nipple. Figure 3a and b show vertical slices through phantoms used for the two series of calculations. The first and second series of phantoms were obtained by asymmetric and symmetric distortions, respectively.

The validity of the above approach was checked by calculating the conversion coefficient, g, using unstructured and structured phantoms with the same boundaries and average compositions. Agreement within 1–5% was found.

RESULTS

Calculations for structured phantoms

Table 1 gives the values of the conversion coefficient, g, calculated for the eight structured phantoms and the 28 kV Mo/Mo spectrum. The corresponding values of g deduced from the tabulations of data based on earlier, simple models of the breast[3,4] are also given. There are significant differences between the two sets of values, which range from 10 to 43%. The differences increase with increasing breast thickness and decreasing glandularity. The reason for the difference between the two sets of values can be understood by reference to Figures 1 and 2. The

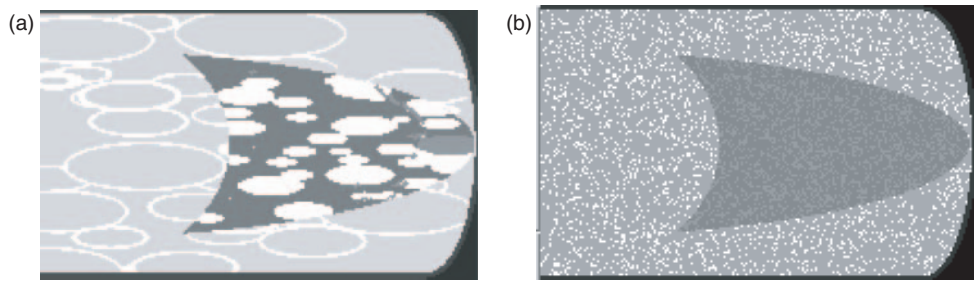

Figure 2. Vertical slices through 50 mm structured (a) and unstructured (b) breast voxel phantoms. Each has a glandularity of 69% in the central region.

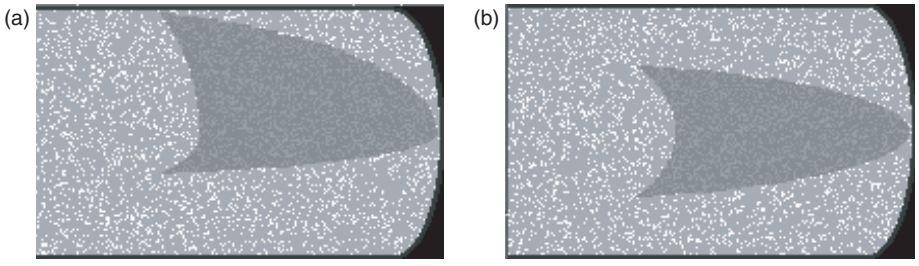

Figure 3. Vertical slices through 50 mm unstructured breast voxel phantoms. Each has a glandularity of 69% in the central region. (a) Left, asymmetric distortion; (b) right, symmetric distortion.

Table 1. Comparison of the conversion factor *g* calculated using a series of high-resolution voxel phantoms with the value calculated using a simple phantom.

Breast thickness (mm)	Breast glandularity (%)	*g*-factor: structured voxel phantom (mGy mGy^{-1})	*g*-factor: simple phantom (mGy mGy^{-1})	Relative difference (%)
40	100	0.177	0.195	−10
40	57	0.202	0.232	−13
40	39	0.216	0.250	−14
50	69	0.140	0.174	−20
60	51	0.114	0.156	−27
60	27	0.122	0.176	−31
80	47	0.0667	0.116	−42
80	25	0.0735	0.130	−43

Note: 28 kV Mo/Mo spectrum, 0.357 mm Al HVL.

average distance of the boundary between the central and surround regions from the breast surface in the structured phantom is greater than the distance of 5 mm used in the simple phantom.

Calculations for unstructured phantoms

Figure 4a and b show the results of the first and second series of calculations for the unstructured voxel phantoms. In both cases, the results are for a 28 kV Mo/Mo spectrum and a 50 mm compressed breast of average glandularity 69%. The results are plotted against the average value, *d*, of the distance from the breast upper surface to the boundary between the central and surround regions of the breast.

The results in Figure 4a show that, depending upon the level of movements and distortion applied, the conversion coefficient, *g*, can be greater than or less than that calculated from the simple model, ranging from 52 to 122% of that value. The two values are equal at a *d*-value of about ~12 mm. Equality for a *d*-value of 5 mm is not expected because of the asymmetrical distribution of

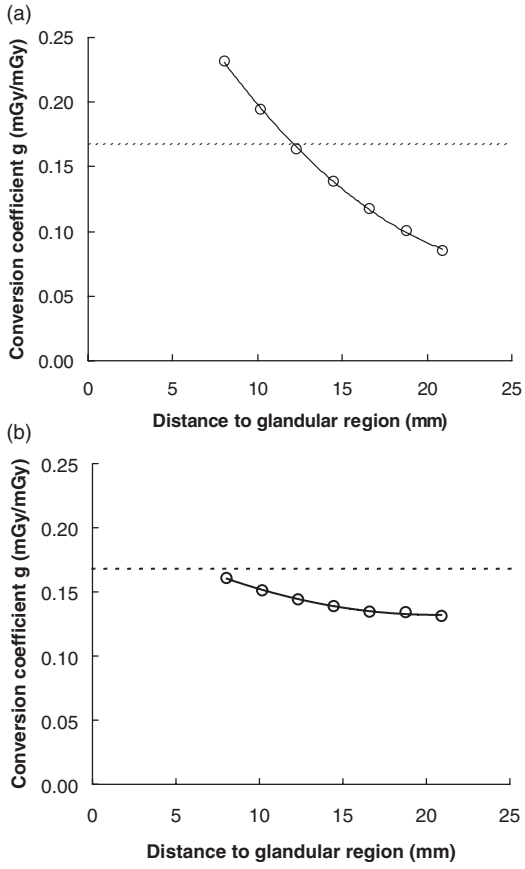

Figure 4. Values of the conversion coefficient *g* calculated using two sets of unstructured phantoms of 50 mm thickness and glandularity 69% in the central region. (a) Phantoms with asymmetric distortion of the central region and (b) phantoms with symmetric distortion of the central region. The horizontal dashed lines give the value of the conversion coefficient deduced from the data and simple breast model of Dance *et al.*[3,4].

glandular tissue, with more tissue being above the midplane of the breast than below. For the symmetrical case, Figure 4b, there is a much smaller variation in the conversion coefficient when compared with the distance *d*, ranging from 75 to 92% of the value using the simple model. This is because of the change in the amount of tissue above and below the midplane is the same. It is noted that the value of the conversion coefficient for a distance *d* of 5 mm is similar to that obtained with the simple model.

DISCUSSION AND CONCLUSIONS

Our Monte Carlo computer model of mammography, incorporating quasi-realistic voxelised models of the breast provides a powerful tool for the study of breast dosimetry. Significant differences have been found between the incident air kerma to MGD conversion coefficients tabulated in the literature and those obtained in this work. For the cases considered, the differences can be as large as 48% and are due to differences in the distribution of the glandular tissue within the breast. These results clearly demonstrate the limitations of the data that are used currently for breast dosimetry.

At present, there are no data available on the actual three-dimensional distributions of glandular tissue within the breast for populations of women. Therefore the methodology cannot yet be used at present to provide better estimates of population dose for screening programmes; it is not suggested at present that the data in current use for the estimation of breast dose be revised.

Data for individual women can in principle be obtained from volume imaging using CT or MR, but the process is difficult and not well established. However, the results of this work can be used to provide better estimates of MGD for individual cases where there is some knowledge of the distribution of glandular tissue within the breast.

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