X-ray absorptiometry of the breast using mammographic exposure factors: application to units featuring automatic beam quality selection

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ABSTRACT. A number of studies have identified the relationship between the visual appearance of high breast density at mammography and an increased risk of breast cancer. Approaches to quantify the amount of glandular tissue within the breast from mammography have so far concentrated on image-based methods. Here, it is proposed that the X-ray parameters automatically selected by the mammography unit can be used to estimate the thickness of glandular tissue overlying the automatic exposure sensor area, provided that the unit can be appropriately calibrated. This is a non-trivial task for modern mammography units that feature automatic beam quality selection, as the number of tube potential and X-ray target/filter combinations used to cover the range of breast sizes and compositions can be large, leading to a potentially unworkable number of curve fits and interpolations. Using appropriate models for the attenuation of the glandular breast in conjunction with a constrained set of physical phantom measurements, it is demonstrated that calibration for X-ray absorptiometry can be achieved despite the large number of possible exposure factor combinations employed by modern mammography units. The main source of error on the estimated glandular tissue thickness using this method is shown to be uncertainty in the measured compressed breast thickness. An additional correction for this source of error is investigated and applied. Initial surveys of glandular thickness for a cohort of women undergoing breast screening are presented.

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A possible link between the mammographic appearance of breast glandular patterns and breast cancer risk was first investigated by Wolfe in 1976 [1], who classified parenchymal patterns into four groups each with increasing amounts of glandular tissue. Wolfe initially found a ratio of 22 in breast cancer incidence between the most glandular and most adipose groups. This ratio was overestimated for several reasons [2], but prompted a number of further studies. Wolfe's original classification was based on the prominence of ductal patterns on xeromammography images. As this technique was replaced by screen-film mammography, this classification was replaced by others based on "breast density" (*i.e.* the proportion and pattern of bright areas in the image denoting glandular tissue). More recent articles [3, 4] have established a significant independent association between breast density and future cancer risk, although at a lower ratio. The field has been reviewed by Harvey and Bovbjerg [5], who identified 12 studies in which quantitative methods of measuring breast density showed a moderate to strong positive association with breast cancer risk. The risk of breast cancer for women with increased breast density in most of these studies is 4–6 times that for women with primarily adipose breasts, a relative risk greater than most traditional risk factors

such as nulliparity and early menarche. Unlike other risk factors, breast density can be influenced by hormonal agents and potentially may be influenced by lifestyle factors such as alcohol intake and diet. Breast density quantification could therefore have a role in risk assessment and prevention decisions, and in decisions to use additional diagnostic tests for women at high risk. Although screen-film mammography is rapidly being replaced by digital mammography, a method that can be applied retrospectively to the large number of images and cancer cases already recorded would be useful to provide evidence for using quantitative measures of the amount of fibroglandular and connective tissue within the breast as a cancer risk indicator. A parallel approach to calibration of a digital mammography unit has been described by Kaufhold et al [6].

Although a number of workers have investigated the automation of breast density classification as an image processing problem (*e.g.* [3, 7–9]), there has been less interest in assessing breast glandularity by means of X-ray absorptiometry. Shepherd et al [10] have demonstrated the use of dual energy X-ray absorptiometry in assessing glandularity in test phantoms and cadaveric breasts, but this requires the use of a separate scanner (normally used for bone mineral assessment) and a small additional radiation dose. The method proposed here, however, uses a calibration of the mammographic X-ray unit used to take the screening mammogram to provide an estimate of breast glandular thickness. The input

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parameters required are already commonly recorded on the X-ray film by modern mammographic units. No additional equipment is required and there is no additional radiation exposure to the subject. This approach has previously been used by our group to estimate breast glandular thickness using the older generation of manual tube potential, fixed target/filter mammography units [11]. It is extended here to modern units that feature automatic beam quality selection.

The proposed method of quantification relies on the use of a well-adjusted automatic exposure control (AEC), a reasonable assumption given the high standard of quality control operated in the UK NHS Breast Screening Programme (NHSBSP), and aims to measure the proportions of adipose and glandular tissue in the column of compressed breast tissue overlying the AEC detector. Modern mammographic units feature sophisticated AECs that are designed to maintain a constant average film optical density over the AEC sensor area, whilst at the same time automatically selecting a suitable X-ray spectrum in response to the sensed compressed breast thickness or to a combination of this and the measured transmission through the breast. Modern units may feature two or three target/filter combinations, any of which may be used over a range of tube potentials. This makes the problem of calibration for X-ray absorptiometry quite challenging.

An approach to calibration at fixed beam quality

Phantom materials mimicking glandular and adipose breast tissue are available in slabs that can be built up into block phantoms with a range of total thickness and glandular content. The phantoms used in this work (CIRS, Norfolk, VA) are based on the formulations of adipose and glandular tissue given by Hammerstein et al [12]. The materials are available in a range of slab thicknesses. The slabs used in this work were in 5, 10 and 20 mm thicknesses, allowing for phantoms up to 90 mm thick with glandular inserts increasing in 5 mm steps. By making a series of automatic exposures over a range of phantom thicknesses and glandular content, the relationship between the phantom thickness, glandular thickness and post-exposure mAs can be established for a given beam spectrum.

One such set of data is illustrated in Figure 1. These results were produced from a Siemens Mammomat 3 (Siemens AG, Erlangen, Germany) at a fixed tube voltage of 28 kVp with a Mo/Mo target/filter combination and are results associated with previous publications [11, 13]. (It is important to note that the glandular thickness definition in the present paper differs from that of percentage glandularity used in [11] and [13] and in other publications on breast dosimetry, in that no constant thickness adipose shield is implied.) The results of Figure 1 can be fitted by exponential functions of mAs as shown [11], but this makes for a somewhat clumsy interpolation of fit coefficients between calibrated compressed breast thicknesses. This approach becomes effectively unworkable for modern units employing perhaps 10 or more different combinations of kVp, target

Figure 1. Calibration using block phantoms between recorded mAs and thickness of breast glandular tissue for a fixed Mo anode/Mo filter mammography unit operated at 28 kVp over a range of compressed breast thicknesses.

and filter that are automatically selected at each exposure.

A series of simplifying approximations can be made by considering the idealised case for a monochromatic beam shown in Figure 2. Here, I_0 is the incident X-ray intensity, I_1 is the intensity reaching the receptor for the 100% adipose case, I_2 is the intensity reaching the receptor when a thickness of glandular tissue, x , is present, T is the compressed breast thickness and μ_A and μ_G are the linear attenuation coefficients for adipose and glandular tissue, respectively. For this monochromatic case:

$$I_1 = I_0 e^{-\mu_A T} \cdot (1 + S) \quad (1)$$

and

$$I_2 = I_0 e^{-\mu_A (T-x)} \cdot e^{-\mu_G x} \cdot (1 + S) \quad (2)$$

where S is the residual scatter fraction at the receptor not removed by the antiscatter grid, which is taken to be approximately the same in both cases. Combining Equations 1 and 2 and rearranging them, the thickness of glandular tissue, x , can be given by:

$$x = \ln(I_1/I_2) / (\mu_G - \mu_A) \quad (3)$$

independent of I_0 and S . Noting that, for a well-adjusted AEC, the post-exposure mAs at a given beam energy will be proportional to the X-ray attenuation at the location of the AEC sensor, x can also be given as:

$$x = \ln(mAs/mAs_0) / (\mu_G - \mu_A) \quad (4)$$

where mAs is the reported exposure with x mm of glandular tissue present and mAs_0 is the mAs for the 100% adipose case with the same compressed breast thickness, T .

Figure 3 shows the data presented in Figure 1 replotted on axes of $\ln(mAs/mAs_0)$ and x . Convincing linear fits through the origin are produced for all thicknesses, T , resulting in a calibration format where only the gradient

Figure 2. Diagrammatic representation of attenuation through a 100% adipose breast of thickness T , compared with that of a breast also of thickness T but containing thickness x of glandular tissue. The initial X-ray intensity in both cases is I_0 , and the attenuated X-ray intensity is I_1 for the 100% adipose case and I_2 for the glandular case.

of the line and mAs_0 value for any given T are needed to estimate x . Departures from the ideal monochromatic case of Figure 2 need to be considered. The good linear fits indicate that beam hardening with increasing glandular thickness, x , is not significant over the range of thickness calibrated. This finding agrees with that of Kaufhold et al [6]. Beam hardening with increasing thickness, T , does however produce the variation in gradient seen in Figure 3, as the quantity $1/(\mu_G - \mu_A)$ increases in value with the increasing effective energy of the beam. The magnitude of this effect can be calculated and used to improve interpolation between calibrated thicknesses.

Figure 4 shows the quantity $1/(\mu_G - \mu_A)$ plotted against photon energy. The values were calculated using the National Institute of Standards and Technology (NIST)

Figure 3. The data of Figure 1 re-plotted on axes of $\ln(mAs/mAs_0)$ and x . Convincing linear fits through the origin are produced for all thicknesses, T , resulting in a calibration format where only the gradient of the line and mAs_0 value for any given T are needed to estimate x .

Figure 4. The quantity $1/(\mu_G - \mu_A)$ plotted against photon energy. The values were calculated using the NIST XCOM program [14] in conjunction with the elemental composition and density data for glandular and adipose tissue given by Hammerstein et al [12].

XCOM (Gaithersburg, MD USA) program [14] in conjunction with the elemental composition and density data for glandular and adipose tissue given by Hammerstein et al [12]. In order to use this relationship as part of the calibration approach, the effective energy of the polychromatic beam reaching the image receptor must be estimated. This was carried out by using a beam spectral simulation program [15] to calculate the half-value layer (HVL) in aluminium at the screen in the screen-film cassette, as the compressed breast thickness is varied. The beam was modelled as 28 kVp Mo target, 30 μ m Mo filter, (630- T) mm of air, 2 mm polymethyl methacrylate (PMMA) compression plate, adipose breast $T=30$ –90 mm plus 4.7 mm PMMA to account for the

Figure 5. The gradients from the linear fits of Figure 3 plotted against photon energy, taken as the effective energy of the polychromatic beam. The dotted line is the theoretical variation in gradient, $1/(\mu_G - \mu_A)$, given in Figure 4, and the solid line through the experimental points is the theoretical curve multiplied by a calibration factor for the experiment.

breast support table, grid interspace material and cover [16], and 3 mm cassette front. The resulting HVL was then matched to the photon energy that would give the same HVL, again using NIST data.

Figure 5 shows the gradients from the linear fits of Figure 3 plotted against an axis of photon energy, where the photon energy is taken as the effective energy calculated above. The dotted line is the theoretical variation in gradient, $1/(\mu_G - \mu_A)$, given in Figure 4, and the solid line through the experimental points is the theoretical curve multiplied by a calibration factor for the experiment, in this case 1.06. This calibration factor accounts for the various approximations linking the monochromatic theoretical model to the polychromatic experimental case, including residual beam hardening from the glandular tissue, and the accuracy of beam modelling in the effective energy calculation. The factor is therefore required if the theoretical curve fit of Figure 4 is to be used as a basis for interpolating between experimental points, a feature that will be more important in the automatic beam selection case described below.

Once the curve of Figure 5 has been established, the only other information required to fully describe the absorptiometric calibration for the mammography unit is the value of mAs_0 for the 100% adipose case, over the range of compressed breast thickness, T . If the natural logarithm of the mAs_0 data from Figure 1 is plotted against T , a convincing straight line is produced that can be used to provide a simple interpolation of mAs_0 for any value of T , shown in Figure 6. This convenient approximation to the monochromatic result in the mammographic energy range has previously been observed by Heine and Behera [17].

Application to mammographic units featuring automatic beam quality selection

The approach described above can be applied to simplify the problem of absorptiometric calibration of mammography units that feature fully automatic beam quality selection. Where the automatic selection is

Figure 6. The natural logarithm of the mAs_0 data from Figure 1 plotted against T , the compressed breast thickness.

restricted to a small number of defined programmes (e.g. the four programmes on the Siemens Mammomat 3000), then the procedure given above can be simply repeated for each. However, for units with more flexible programming (e.g. the Lorad M-IV, Hologic Inc., Bedford, MA, USA; the GE DMR, GE Healthcare, Chalfont St Giles, Bucks, UK), the kVp and/or target and beam filter may change at any stage during the block calibration process, either as the total phantom thickness, T , is increased or as the thickness of glandular tissue, x , is increased. This makes it difficult to obtain the full range of measurements ideally required for a reliable calibration.

As a guide and to limit the number of combinations that need to be measured, a previously established relationship between compressed breast thickness and percentage glandularity associated with the age group routinely called for breast screening in the UK can be used [11, 13]. The fit to mean percentage glandularity with compressed breast thickness for a 50–64-years-old age group from reference [13] (which includes a 5 mm adipose shield at all compressed breast thicknesses) was converted to glandular thickness without the assumption of an adipose shield, and without the constraint of 100% glandularity at 20 mm compressed breast thickness. This is shown plotted against compressed breast thickness in Figure 7. The curve can be used as the basis for a set of phantom thickness and glandular thickness combinations that will cover the average range found in breast screening. Three glandular thicknesses bracketing the expected value were used at each phantom thickness to reduce the error on the fitted gradient of glandular thickness, x , against $\ln(mAs/mAs_0)$. 100% adipose was included to give the mAs_0 value in each case. The choice of four points for each line fit is arbitrary, but it was found in practice that the experimental error on the gradient increased rapidly with fewer points than this.

Calibration measurements were carried out on three examples of the Lorad M-IV unit operated in “auto filter” mode. On this setting the unit selects kVp over the range 25–32 kVp and changes the filter from Mo to Rh depending on the compressed breast thickness and attenuation sensed by the AEC. In cases where the

Figure 7. Expected average values of glandular tissue thickness plotted against compressed breast thickness for 50–64-years-old age range. Data derived from the surveys in references [11] and [13].

machine varied the kVp and/or filter at a single compressed breast thickness as the glandular thickness was increased, additional measurements using fixed kVp and filter ("auto time" mode) were used to obtain the four calibration points at that compressed breast thickness. Additional measurements with 100% adipose phantoms were also made to establish the mAs_0 values needed to cover the range of thickness, kVp and filter combinations programmed into the unit. Each full calibration took approximately 3 h to perform.

An important source of possible error in the calibration is the accuracy of the compressed breast thickness measurement given by the unit. The value given will depend on the compression force applied and on the amount of flexing the compression plate allows. For this work on rigid phantoms, it was thought reasonable to drive the plate manually so that it touched the phantom surface, but did not apply a compression force that would then deform the plate. On the units tested this

approach gave exactly the calibrated thickness, implying that this is the way the units had been set up.

Calibration results

Figure 8 shows the calibration results for three Lorad M-IV mammography units. The figure shows the fitted gradient of glandular thickness, x , with $\ln(mAs/mAs_0)$, plotted against effective beam energy. The effective beam energy now takes account of the different kVp/target/filter combinations automatically selected by the unit, as well as beam hardening with phantom thickness, and therefore extends over a larger energy range than the simple single kVp case previously shown in Figure 5. In Figure 8 the solid line is the theoretical curve for $1/(\mu_G - \mu_A)$ derived above fitted to the experimental points with the calibration factors shown. All are close to unity. The errors in the experimental points are calculated as the root mean square deviation (RMSD) in

Figure 8. The calibration results for three Lorad M-IV mammography units (a–c). The fitted gradient for the relationship between $\ln(mAs/mAs_0)$ and x is plotted against the effective energy of the beam reaching the image receptor. The solid line in each case is the theoretical curve fitted to the experimental points with the calibration factors shown. The error bars represent ± 2 root mean square deviation in the fitted gradient.

Figure 9. An example of the fits of $\ln(mAs_0)$ against compressed breast thickness for the various kVp and filter combinations encountered during the calibration process. This example corresponds with the calibration of Figure 8a and shows fits for (L–R) 25 kVp Mo/Mo, 26 kVp Mo/Mo, 27 kVp Mo/Mo, 28 kVp Mo/Mo, 29 kVp Mo/Mo, 29 kVp Mo/Rh, 30 kVp Mo/Rh, 31 kVp Mo/Rh and 32 kVp Mo/Rh.

the gradient due to the spread of experimental values of $\ln(mAs/mAs_0)$ about the fitted gradient. The error in the thickness measurement of the rigid blocks is taken as zero. The error bars indicate ± 2 RMSD. With the calibration factors applied, the experimental points are a good fit to the theoretical relationship for all three units. Figure 9 shows an example of the corresponding fits of $\ln(mAs_0)$ against compressed breast thickness for the various kVp and filter combinations encountered during the calibration process. Thus, for any given value of compressed breast thickness and beam energy, the value of mAs_0 for the 100% adipose case can be estimated and

substituted into Equation 4 together with the actual recorded mAs (adjusted if necessary for any differences in fine density setting between calibration and clinical use) and the calibrated value of $1/(\mu_G - \mu_A)$ derived from the fit of Figure 8, to produce an estimate of x , the thickness of glandular tissue within the breast.

Effect of error in calibration curve fits

To investigate the magnitude of errors arising from the various curve fits in the calibration method, the original phantom exposure mAs values ($n=122$) for the three mammography units were run back through the calibration process and the differences between the estimated and true glandular thicknesses calculated. The resulting errors were found to be normally distributed with an RMSD of 1.0 mm.

Effect of error in compressed breast thickness estimation

Owing to possible inaccuracy in the indicator calibration, and the inevitable difference in the mechanics of compression between a rigid reference block and real breasts, it would be expected that error in compressed breast thickness estimation could have a strong influence on the estimated glandular thickness. Kaufold et al [6] analysed the effect of this source of error on estimation of percentage glandularity for a calibration of a digital mammography unit, referencing the work of Burch and Law [18] and Highnam and Brady [19] who estimated a typical error on compressed breast thickness of ± 2 mm. A similar approach is used here.

Table 1 shows a summary of the results obtained by calculation of the errors on glandular thickness estimation using an illustrative error of ± 2 mm on true thicknesses ranging from 20 mm to 90 mm. At each

Table 1. Fractional errors on glandular thickness and percentage glandularity calculated for a ± 2 mm error in compressed breast thickness^a

True breast thickness (mm)	Expected glandular thickness (mm)	Expected % glandularity	Breast thickness error (mm)	Measured breast thickness (mm)	kV and target/filter combination	Glandular thickness (mm)		% Glandularity	
						Estimate (mm)	Fractional error	Estimate (%)	Fractional error
20	11.5	57.5	+2	22	25 Mo/Mo	8.0	-0.31	36.3	-0.37
30	15.6	52.0	+2	32	25 Mo/Mo	12.0	-0.23	37.4	-0.28
40	15.2	38.0	+2	42	26 Mo/Mo	11.3	-0.26	26.9	-0.29
50	12.6	25.2	+2	52	28 Mo/Mo	9.1	-0.28	17.4	-0.31
60	9.5	15.8	+2	62	28 Mo/Mo	5.8	-0.39	9.4	-0.41
70	7.0	10.0	+2	72	29 Mo/Mo	3.3	-0.52	4.6	-0.54
80	5.4	6.8	+2	82	30 Mo/Rh	1.2	-0.78	1.5	-0.78
90	4.5	5.0	+2	92	32 Mo/Rh	-0.1	-1.02	-0.1	-1.02
20	11.5	57.5	-2	18	25 Mo/Mo	15.0	0.31	83.4	0.45
30	15.6	52.0	-2	28	25 Mo/Mo	19.2	0.23	68.7	0.32
40	15.2	38.0	-2	38	26 Mo/Mo	19.1	0.26	50.3	0.32
50	12.6	25.2	-2	48	28 Mo/Mo	16.1	0.28	33.6	0.33
60	9.5	15.8	-2	58	28 Mo/Mo	13.3	0.39	22.9	0.44
70	7.0	10.0	-2	68	29 Mo/Mo	10.6	0.52	15.7	0.57
80	5.4	6.8	-2	78	30 Mo/Rh	9.6	0.78	12.3	0.82
90	4.5	5.0	-2	88	32 Mo/Rh	9.1	1.02	10.3	1.06

^aThe illustrative error was used over the range of compressed breast thickness 20–90 mm. The expected glandular thickness from Figure 7 is used [13] and the beam quality selection matches that produced in the calibration experiments.

compressed breast thickness, the average glandular thickness from Figure 7 was used as a typical value, and calibration values for one of the units reported above (that of Figure 8a) were used to calculate the glandular thickness at the beam quality automatically selected for the closest matching exposure from the calibration set. The estimated value is expressed in two ways: as glandular thickness in mm and as percentage glandularity (equal to glandular thickness divided by measured compressed breast thickness $\times 100$). The errors are shown as fractional errors to avoid confusion with the percentage glandularity.

Two points are immediately apparent from Table 1. Firstly, the fractional errors can be large, especially at large compressed breast thicknesses where the typical glandular thickness is small. Secondly, the fractional errors in the percentage glandularity are consistently larger than those for the glandular thickness. This might be expected, as the error on the compressed breast thickness also appears in the denominator of this value. As a result of this finding, it was decided to use only glandular thickness in mm as the reported value.

Kaufold et al [6] report a standard deviation of $\pm 7\%$ for their estimate of percentage breast glandularity for a 40 mm compressed breast with a true 50% glandularity, by treating the ± 2 mm error in compressed breast thickness as a uniformly distributed variable with an equal probability of any value between -2 mm and $+2$ mm. They also show a mapping of true to estimated percentage glandularity for a true 40 mm breast for the extreme error values of -2 mm and $+2$ mm. From this latter diagram it is possible to estimate their fractional errors in a true 40 mm breast with a true 45% glandularity, allowing direct comparison with the values for the 40 mm breast given in Table 1. Their resulting fractional errors of -0.30 for a $+2$ mm thickness error and 0.27 for a -2 mm thickness error are similar to the equivalent values in Table 1. This indicates that the two

methods are comparable in their sensitivity to thickness errors.

Effect of error in recorded post-exposure mAs

A further predictable source of error in the glandular thickness estimation is the variation in post-exposure mAs due to variation in attenuation between individual film-screen cassettes. In mammography, the AEC sensor lies behind the cassette so variation in attenuation will be reflected in the mAs recorded. The quality standard in place for the NHSBSP is that the variation between cassettes should result in a maximum variation of no greater than $\pm 5\%$ from the mean mAs for the batch of cassettes [20].

Table 2 shows a summary of the results obtained by calculation of the errors in glandular thickness estimation using an illustrative error of $\pm 5\%$ on the true mAs. The layout and methodology is similar to that of Table 1. As might be expected, the fractional errors in glandular thickness and percentage glandularity are the same in this case. The overall size of the errors from this source are smaller than those from errors in the compressed breast thickness above, but are still significant where the expected value of glandular thickness is small.

Reduction of systematic compressed breast thickness errors

Owing to the known high sensitivity of the glandular thickness estimation to errors in compressed breast thickness, and the known incompatibility between the thickness calibration of the estimation method (zero compression force, rigid block phantom) and clinical mammography (large compression force plus flexing/tilting compression plate), this aspect was investigated more fully. Previous work indicates that under clinical

Table 2. Fractional errors in glandular thickness and percentage glandularity calculated for a $\pm 5\%$ error in recorded mAs^a

True breast thickness (mm)	Expected glandular thickness (mm)	Expected % glandularity	mAs error (%)	kV and target/filter combination	Glandular thickness (mm)		% Glandularity	
					Estimate (mm)	Fractional error	Estimate (%)	Fractional error
20	11.5	57.5	+5	25 Mo/Mo	12.8	0.11	63.8	0.11
30	15.6	52.0	+5	25 Mo/Mo	16.9	0.08	56.3	0.08
40	15.2	38.0	+5	26 Mo/Mo	16.6	0.09	41.5	0.09
50	12.6	25.2	+5	28 Mo/Mo	14.1	0.12	28.2	0.12
60	9.5	15.8	+5	28 Mo/Mo	11.1	0.17	18.5	0.17
70	7.0	10.0	+5	29 Mo/Mo	8.7	0.25	12.5	0.25
80	5.4	6.8	+5	30 Mo/Rh	7.5	0.39	9.4	0.39
90	4.5	5.0	+5	32 Mo/Rh	6.9	0.54	7.7	0.54
20	11.5	57.5	-5	25 Mo/Mo	10.2	-0.11	50.9	-0.11
30	15.6	52.0	-5	25 Mo/Mo	14.2	-0.09	47.5	-0.09
40	15.2	38.0	-5	26 Mo/Mo	13.7	-0.10	34.4	-0.10
50	12.6	25.2	-5	28 Mo/Mo	11.0	-0.13	22.0	-0.13
60	9.5	15.8	-5	28 Mo/Mo	7.9	-0.17	13.1	-0.17
70	7.0	10.0	-5	29 Mo/Mo	5.2	-0.26	7.4	-0.26
80	5.4	6.8	-5	30 Mo/Rh	3.1	-0.41	3.9	-0.41
90	4.5	5.0	-5	32 Mo/Rh	2.0	-0.57	2.2	-0.57

^aThe illustrative error was used over the range of compressed breast thickness 20–90 mm. The expected glandular thickness from Figure 7 is used [13] and the beam quality selection matches that produced in the calibration experiments.

conditions the indicated breast thickness may considerably underestimate the true breast thickness [21–23].

A number of approaches were tried to produce a thickness correction to the value recorded by the X-ray unit, using the values of AEC position and compression force also recorded for each exposure. The best approach proved to be the use of a silicone breast prosthesis placed on rigid spacer blocks to give a test object with a realistic compliance and contact area on the compression plate. Measurements were made for a range of compression force and AEC position, with the phantom centred on the selected AEC position each time. The form of the correction was based on that given by Mawdsley et al [23], with the force and plate tilting terms fitted to the experimental measurements, and the offset term fitted to the baseline glandular thickness survey results, but being a similar value as that given by Mawdsley et al.

Baseline measurements of glandular tissue thickness

Figure 10 shows the results of an initial calculation of glandular tissue thickness as a frequency plot for 3571 exposures recorded on one of the units calibrated above. These data form the basis of a survey of predominantly normal women being used in retrospective epidemiological studies of cancer risk and breast density.

Figure 11 shows a re-working of the data from Figure 10, limited to lateral oblique views for age 50 to 64 and analysed as average glandular thickness as a function of compressed breast thickness. Superimposed on the data points is the curve of Figure 7, which was derived from two previous surveys [11, 13] for ages 50 to 64. The current results show a reasonable agreement with the trends of the previous surveys, bearing in mind that the previous survey line is derived from a curve fit to percentage glandularity rather than to glandular thickness directly.

Figure 10. A frequency plot of glandular tissue thickness for 3571 exposures recorded on one of the calibrated mammography units.

Figure 11. The data from Figure 10 restricted to age 50–64, lateral oblique only, and re-plotted as average glandular thickness against compressed breast thickness (points). The error bars show ± 1 standard error in the mean. Superimposed is the curve from Figure 7, which was derived from previous surveys for this view and age range [11, 13].

Conclusion

Approximately 1.8 million women have X-ray mammography within the UK NHSBSP each year; for a significant proportion of these, the exposure parameters related to each examination are routinely recorded on the X-ray film. This information represents a large potential source of data on breast composition and its relationship to breast cancer risk. A problem in employing these data is the difficulty in calibrating modern mammography units that feature automatic beam quality selection to perform X-ray absorptiometry. Using appropriate models for the attenuation of the glandular breast in conjunction with a constrained number of physical phantom measurements, it has been demonstrated that such a calibration can be achieved despite the large number of possible exposure factor combinations used by modern mammography units. The main problem with the method is the dependence on an accurate measurement of the compressed breast thickness.

Although the future of mammography is undeniably digital, it is important that the relationship between breast glandularity and breast cancer risk is investigated with the current generation of mammography units using the large number of cases already recorded. In this work, the glandular thickness in a column of tissue overlying the sensor of the automatic exposure control is estimated and proposed as a quantitative measure of breast composition for risk prediction. This quantity represents an intermediate step between the visual assessment of breast density used in much of the existing work on risk estimation and the full volumetric measurement of glandular content that can be achieved with digital mammography. If a predictive relationship can be established with the present method, then the case for a move to routine and systematic risk assessment using full volumetric measurement of glandular tissue from digital mammography will be strengthened.

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