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Design of a novel phase contrast x-ray imaging system for mammography

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Abstract

It is hoped that x-ray phase contrast imaging (XPCi) will provide a generational improvement in the effectiveness of mammography. XPCi is sensitive to the refraction which x-rays undergo as a result of the variation in x-ray propagation speeds within an object. XPCi is, however, seldom used in clinical applications owing mainly to a lack of suitable systems. The radiation physics group at UCL has previously designed and built an XPCi system sensitive to phase gradients in one dimension for application in security inspection. We present here the design methodology and final design of a prototype XPCi system sensitive to phase gradients in two directions for use in mammography. The technique makes efficient use of the flux available from a laboratory x-ray source, thus making it suitable for clinical use.

1. Introduction

It is hoped that x-ray phase contrast imaging (XPCi) will provide a generational improvement in the effectiveness of mammography (Lewis 2004). An XPCi technique employing laboratory sources, suitable for mammography, was suggested by Olivo and Speller (2007a), (2008b). This technique is known as coded aperture XPCi (CAXPCi) and has since been under continuous development within the radiation physics group at UCL (see Olivo *et al* (2009a), Olivo and Speller (2007a), (2007b) for example). This technique has been demonstrated experimentally and validated theoretically in the aforementioned references principally for CAXPCi systems sensitive to phase gradients in one dimension. We are now building a pre-prototype CAXPCi system sensitive to phase gradients in two dimensions based on the initial work of Olivo *et al* (2009a). This system will be used to assess the efficacy of the technique using *in vitro* human breast tissue samples.

To our knowledge, the only *in vivo* mammography program currently in progress is in Trieste, Italy, using the SYRMEP beam line (Castelli *et al* 2007, Dreossi *et al* 2008). This program has provided mammograms of improved contrast and detail visibility compared with conventional mammography. A clinical trial is in progress but results are yet to be released. The images are acquired using a technique known as free space propagation (FSP) XPCi which requires an x-ray beam of high spatial coherence but substantially limiting the emitted flux. Such a beam may be obtained using synchrotron radiation, as employed in the Trieste program, making it impractical for clinical screening. This technique has been applied using microfocal or strongly apertured laboratory sources (Wilkins *et al* 1996). This, however, results in a prohibitively long exposure time owing to the low flux available from such sources.

A FSP XPCi system was developed using conventional x-ray sources with a nominal focal spot size of $100\text{ }\mu\text{m}$ (Morita *et al* 2008, Tanaka *et al* 2009). In a clinical trial encompassing 3835 examinations, the system was found to not provide a statistically significant difference in recall rates and cancer detection rates when compared with conventional film screening (Morita *et al* 2008). It was, however, reported that the system resulted in superior depiction of abnormalities. To the best of our knowledge, this system has experienced very limited clinical uptake. The main reason for this is that a source focal spot size of $100\text{ }\mu\text{m}$ is necessary to achieve sufficient source flux yet the contrast developed by the FSP method reduces as the source size increases (Arfelli *et al* 1998).

Other methods of performing XPCi using laboratory sources suffer from the same problem as the FSP method. In particular, the Talbot interferometric technique (Momose *et al* 2006) works only with microfocal sources and the Talbot-Lau interferometer (Pfeiffer *et al* 2006) employs an aperture in front of a standard laboratory source which creates an array of sources each emitting a beam of high spatial coherence. Both techniques result in exposure times which are too long for clinical use.

The proposed CAXPCi system works by partially illuminating detector pixels to emphasize the effect of refracted photons on the detector pixel signals. We do not describe the technique in detail; instead, the reader is referred to a recent publication (Olivo and Speller 2008a). A schematic diagram of a CAXPCi system, sensitive to phase gradients in one direction, is shown in figure 1. When a refracting object is placed on the detector side of the sample apertures, some photons which previously did not reach a detector pixel will do so and vice versa. This creates phase contrast in the detected image. An image sensitive to phase gradients in two directions could be obtained by taking two exposures with the one-dimensional system, each for a different orientation of the object relative to the apertures. Such an approach has been suggested by Kottler *et al* (2007); however, this is not practical for clinical application due to the technical difficulties and time associated with rotating the imaging system. The proposed system will instead employ apertures with a two-dimensional transmission profile. In the remainder of the paper, we consider the possible designs of such two-dimensional apertures and discuss the trade-offs leading to the chosen aperture type. We then model the imaging of breast tumours and calcification in order to optimize the system parameters.

2. Two-dimensional aperture designs

Before considering the proposed aperture designs, it is important to explain the concept of illuminated pixel fraction. As previously explained, CAXPCi works by illuminating the edge of a pixel, leaving some of the pixel, not covered by the detector aperture, in shadow. When a point source is employed, the illuminated fraction is defined, in the regime of geometrical optics, as the proportion of the total exposed pixel area illuminated by x-rays in the absence of an object. This is demonstrated in figure 2 which shows an x-ray beam incident upon the exposed part of a pixel. In this case the illuminated fraction is the proportion of the exposed part of the pixel which is illuminated by x-rays and is given by a/b . This definition does not allow for the effects of diffraction and a finite sized source. The most general definition of the illuminated fraction is the ratio of the pixel signal to the maximum obtainable pixel signal which is generally achieved when the sample and detector apertures are perfectly aligned with respect to the centre of the source distribution.

Throughout this paper, we describe the aperture shape by a single cell which is used to create an entire mask by periodic repetition as shown in figure 3. In order to be sensitive to phase gradients in two orthogonal directions, an aperture must be composed of two or more apertures arranged orthogonally. The sample and detector apertures must also be able to be

shifted relative to one another in order to achieve a sufficiently low, variable, fraction of illumination. This requirement impacts strongly on the aperture shape as a single pair of sample and detector apertures must allow x-rays to be incident on one detector pixel only. In practice, sources of finite size prevent this from being achieved; however, it is important that the vast majority of x-rays admitted by a particular sample/detector aperture pair are incident upon a single detector pixel. These constraints allow three main aperture shapes which are illustrated in figure 4.

The three apertures in figure 4 have much in common. Aperture 2 was however determined to be unsuitable as it does not permit a sufficiently low illuminated fraction to be achieved. To understand this, consider viewing the imaging system from the location of the source. A low illuminated pixel fraction will be observed when only a small fraction of the transmitting region of the sample aperture is projected onto the transmitting region of the detector aperture. The constraint of keeping both transmitting regions within the projection of a single pixel means that even when maximally shifted, the vertical part of the detector aperture will overlap with the horizontal part of the sample aperture and vice versa as shown in the lower row of images in figure 4. Aperture 2 was thus discarded from further investigation.

Apertures 1 and 3 are very similar with the main difference being that aperture 3 is marginally more sensitive to sample phase gradients at the cost of reduced signal compared with aperture 1. Due to this similarity, the next phase of analysis was performed only for aperture 1. Two metrics were used to refine the design of aperture 1. The first of these is the total x-ray flux which an aperture will admit which may be measured by the area of the transmitting region of aperture. For a single cell of aperture 1, the proportion of the cell which is transmitting, and thus the proportion of incident flux transmitted, is

$$\Phi_{\text{cell}} = (1 - \eta_2)^2 - (1 - \eta_2 - \eta_1)^2. \quad (1)$$

The potential sensitivity to phase gradients in the sample can be approximated by the length of aperture edge e_1 in figure 4 given by

$$\sigma_{\text{cell}} = 1 - \eta_2 - \eta_1. \quad (2)$$

In order to preserve sufficient flux and sensitivity, it was decided to restrict η_1 and η_2 to

$$\begin{aligned} 0.1 &\leq \eta_1 \leq 1/3 \\ \eta_1 &\leq \eta_2 \leq 1 - 2\eta_1. \end{aligned} \quad (3)$$

Figure 5 shows intensity plots of σ_{cell} and Φ_{cell} over the range of values of η_1 and η_2 defined in (3). Also, annotated on these plots is an approximate region of values of η_1 and η_2 chosen on the basis of the plots of σ_{cell} and Φ_{cell} . This region was subsequently refined using a more detailed analysis of the illuminated pixel fraction taking into account the source dimension and magnification.

The illuminated pixel fraction may be found by calculating the diffracted field incident upon the detector apertures due to the x-ray source having finite spatial and spectral widths. In general, such a field may be calculated for a monochromatic point source, the intensity of which is then convolved with the effective source distribution. The source distribution is assumed to be Gaussian and is described by the effective source full width at half maximum (FWHM), $W_{\text{eff}} = W(M - 1)$, where W is the actual source FWHM and M is the system magnification defined as $(z_{\text{so}} + z_{\text{od}})/z_{\text{so}}$. z_{so} and z_{od} are the source to sample and sample to detector distances as depicted in figure 1. The usage of W_{eff} is justified rigorously in a

recent paper (Munro *et al* 2010). Our simulations show that, for the purposes of calculating the fraction of illumination, the illuminated fraction will be approximately equal for different combinations of W and M providing that they result in the same value of W_{eff} . Furthermore, we found that simulations performed at the average source photon energy accurately predict the illuminated pixel fraction of the broadband source.

Figure 6 shows intensity plots of the minimum fraction of illumination for a selection of values of W_{eff} , η_1 and η_2 for the three possible pixel dimensions. The pixel dimensions of 50, 85 and 100 μm were chosen according to the pixels of the candidate detectors as discussed in section 3. These results show that the pixel dimension of 50 μm is impractical as it is impossible to achieve a sufficiently low illuminated pixel fraction. These results are combined with those of later sections to determine the final system design.

3. Choice of detector

As the objective of this project is to design a prototype system, it was decided that only off-the-shelf components should be used. As a result, it was rapidly decided that flat panel arrays are the most suited devices for the investigation foreseen by this project, primarily because they are the current standard in mammography. Moreover, we decided that both a direct conversion and indirect conversion detector should be evaluated. The only practical direct conversion option was the ANRAD a-Se (SMAM) flat panel detector, featuring a pixel size of 85 μm . As far as the indirect conversion method is concerned, the choice is much wider. However, in order to provide a fair comparison with the 85 m pixel ANRAD detector, a flat panel with similar spatial resolution characteristics is required; otherwise, we would be comparing detectors with a wildly different spatial resolution rather than detectors with a similar resolution but based on different technologies. An analysis of the spatial resolution characteristics of commercially available indirect conversion flat panels demonstrated that the 50 μm pixel HAMAMATSU (C9732DK passive pixel CMOS flat panel with directly deposited structured CsI) is probably the closest match in terms of spatial resolution. The 100 μm pixel dimension previously mentioned is obtained by grouping pairs of pixels on the HAMAMATSU detector.

For our application, an important indicator of detector performance is the so-called pixel spill-out. The spill-out describes the amount of signal that a pixel registers as a result of one of its neighbours being illuminated by x-rays. The spill-out of both detectors is derived in detail in a recent paper (Olivo *et al* 2009b) and have been reproduced in table 1. This data will be used principally in section 4 where the CAXPCi image of tumours and calcifications is calculated.

4. Tissue modelling

Data governing the interaction of x-rays with human breast tissue were obtained from three primary sources: ICRU report number 46 (ICRU 1991), Johns and Yaffe (1987) and Al-Bahri and Spyrou (1998). Partial overlap exists between the results of each of these publications. The ICRU publication contains the most exhaustive set of data, giving the elemental compositions, mass density, electron density and mass attenuation coefficients for a variety of human tissue types. Of particular interest are the data for breast tissue. Three types of breast tissue from this report are considered: calcifications and two types of whole breast tissue, intended to represent the age-dependent changes in the whole female breast. Tumour tissue is a notable omission from these data and so the publication by Johns and Yaffe (1987) is used to obtain the linear attenuation coefficient for tumour tissue (infiltrating duct carcinoma) and Al-Bahri and Spyrou (1998) is consulted for the electron density of tumour tissue. The simulations in this paper assume that calcifications and tumours are

accommodated in tissue, the range of which is represented by the two ‘whole breast’ samples mentioned previously.

Experiments have shown (Olivo and Speller 2006) that modelling the CAXPCi system at the source’s average photon energy gives a good prediction of the image due to a polychromatic source. We consider here a photon energy of 17.5 keV, the dominant characteristic peak encountered when employing a Mo anode for tube potentials commonly employed in mammography. The ICRU mass attenuation coefficients had to be interpolated to this photon energy using cubic interpolation. The lowest photon energy considered by Johns and Yaffe was 20 keV and so the linear attenuation coefficient had to be extrapolated, again using cubic extrapolation. Carroll *et al* (1994) measured the mass attenuation coefficients of tumour tissue and healthy tissue. They found that for photon energies between 14.15 and 18 keV, the linear attenuation coefficient of tumour tissue was on average 10.9% higher than that of healthy tissue. They also noted that some overlap existed. The data of Johns and Yaffe (1987) lead to a significantly greater difference between tumour and healthy tissue. We have chosen to use the data of Johns and Yaffe as Carroll *et al* do not give absolute figures for the attenuation coefficients. Furthermore, this choice leads to a more conservative estimate of the improvement of CAXPCi versus conventional absorption imaging.

It is customary to define the refractive index at x-ray energies as $n = 1 - \delta + i\beta$ where β represents absorption and δ the phase term. β is calculated from the linear attenuation coefficient according to $\beta = \mu/(2k)$ where $k = 2\pi/\lambda$ is the wave number and λ is the wavelength. δ is calculated each tissue type according to (Vo-Dinh 2003)

$$\delta = 4.49 \times 10^{-16} \lambda^2 N_e, \quad (4)$$

where N_e is the electron density of the tissue. Table 2 summarizes the raw data and the calculated values of δ and β for the tissue types considered.

Note that when considering small tumour spicules and calcifications, one should ideally consider tumours to be embedded in distinct tissue types rather than a combination of glandular and fat as considered in this paper. The results presented in this paper should thus be considered average over these two types of embedding tissue. We do not expect this to affect the salient features of the simulations used to design the CAXPCi system.

The data in table 2 were used to simulate the imaging of tumour tissue and calcifications by the proposed CAXPCi system using a wave optical model of the CAXPCi system published recently (Munro *et al* 2010). The proposed two-dimensional apertures (see apertures 1 and 3 in figure 4) consist of two orthogonal apertures which are each sensitive to phase gradients in one direction only. Because of this we have modelled a one-dimensional imaging system in order to optimize the design of the system and determine which detector to employ. We consider the spicules of tumours to be approximated by cylinders of varying radius. Although calcifications are not cylindrical, they too were modelled as cylinders. In doing this we are tacitly assuming that the calcifications can be approximated by spheres and that in a one-dimensional CAXPCi system, spheres are well approximated by cylinders. This is reasonable as any smooth curve may be approximated locally by a circle of appropriate centre and radius. Using the one-dimensional approximation significantly reduces the computational cost of the simulations. The simulated tumours and calcifications were assumed to be embedded in a 3.5 mm thick slice of healthy tissue as is the expected thickness of samples to be used during *in vitro* testing of the system. The image profiles were considered representative of the full image detail and were therefore used to evaluate

the contrast. Thus, in what follows we sometimes refer to an image when strictly we should perhaps refer simply to a profile.

The simulations were performed using the imaging system depicted in figure 1 and for every combination of the parameter ranges listed in table 3. We first present CAXPCi system simulated results to demonstrate the nature of the images obtained by the system. Figure 7 shows the CAXPCi, conventional absorption and FSP signals for a fibre of radius 0.1 mm composed of tumour tissue. All of the simulation parameters are given in the caption of figure 7 and so are not repeated here. The conventional absorption image was obtained with the sample assumed to be positioned adjacent to the detector. The FSP image was obtained under the same conditions as the CAXPCi image, except that the sample and detector apertures were not present. For each plot in figure 7, the sample was scanned in 1 μm increments. This technique, also known as dithering, allows a much higher spatial resolution to be obtained than would otherwise be permitted by the pixel dimension. Dithering would be impractical in a clinical context but we have simulated it in order to assess both the extreme and expected performance of the system. In particular, the images used to construct the dithered image will, in general, exhibit differing contrasts. From these images we can obtain the best and worst case as well as the expected performance.

One can see from the plots in figure 7 that the edges of the fibre are enhanced even when employing the FSP method but are enhanced much more significantly in the CAXPCi image. This demonstrates that the FSP method provides very little advantage over conventional absorption and perhaps why the uptake of the system developed by Konica–Minolta has been limited. Also, evident from the plots is that pixel spill-out leads to the introduction of spurious ripples in the CAXPCi signal as well as a reduction in contrast, albeit very minor in the example displayed. Figure 8 shows a plot of the signal acquired when calcifications are considered instead of tumour tissue. Comparison of figures 8 and 7 reveals that tumour tissue results in an image dominated by the phase of the tumour whilst the calcification image is dominated by absorption with some edge enhancement.

We use contrast as a means of objectively comparing CAXPCi with conventional absorption-based imaging. For this purpose we define the contrast as $\gamma = (I_{\text{max}} - I_{\text{min}}) / (I_{\text{max}} + I_{\text{min}})$ where I_{max} and I_{min} are the maximum and minimum image intensities, respectively. The improvement in contrast resulting from CAXPCi may then be expressed as the ratio of contrasts observed using CAXPCi and conventional absorption-based imaging and FSP imaging. Figure 9 shows plots of such a ratio for a typical set of imaging parameters which may be employed in practice in order to demonstrate the potential of the technique. As expected, the improvement in contrast offered by CAXPCi increases for smaller radii. In practice visual clutter, resulting from the projection of a three-dimensional breast onto a two-dimensional image, and noise will prevent such small details as considered in figure 9 from being detected. Also, of note is that the improvement in contrast is much greater in the case tumours compared with calcifications. This is largely because calcifications are highly absorbing and, as shown in figure 8, result in an CAXPCi image dominated by absorption. Figure 10 shows the improvement in contrast offered by CAXPCi compared with conventional absorption imaging for a typical set of imaging system parameters as a function of source FWHM. We chose a lesion/calcification radius of 100 μm as details of this size are generally not visible in conventional mammography. The trends demonstrated in the plots are consistent with other choices of imaging system parameters such as M , fraction of illumination and detail radius. The plots demonstrate that the contrast improvement is greater for the 85 μm pixel than in the case of the 100 μm pixel. This is to be expected as the 85 μm pixel system has a greater number of slightly narrower x-ray beams as compared with the 100 μm pixel for same fraction of illumination. As demonstrated previously by Olivo and Speller (2008a), CAXPCi contrast increases as the

directly illuminated part of a pixel is reduced. If two CAXPCi systems employ different pixel dimensions but the same fraction of illumination then, other things being equal, both systems detect the same total number of photons. However, in absolute terms, the system with the smallest pixel dimension illuminates a smaller region of each pixel. Figure 10 also shows that the 85 μm pixel system suffers more from pixel spill-out than the 100 μm system as is expected from table 1. It is important to note that for a source of practical FWHM, the 85 μm system still has a greater or comparable contrast improvement compared with the 100 μm system even when pixel spill-out is considered. A curious feature of the plots in figure 10 is that when calcifications are considered, contrast improvement is marginally greater when pixel spill-out is considered for source FWHM of 50 and 75 μm . This is partially a numerical artifact as the loss in contrast in the conventional image, owing to pixel spill-out, is greater than the corresponding loss in the CAXPCi system. This does not imply that the CAXPCi contrast increased when pixel spill-out was considered, only that it was degraded to a lesser extent than the conventional image. Finally, figure 10 demonstrates, as expected, that the contrast improvement of CAXPCi reduces as the source FWHM increases.

Figures 11-14 show plots of the improvement in contrast offered by CAXPCi over conventional absorption-based imaging for a range of radii and fractions of illumination as a function of magnification. Five magnification values were considered: 1.33, 1.29, 1.25, 1.21 and 1.18. Figure 11 considers a 100 μm pixel and calcification, figure 12 a 100 μm pixel and tumour tissue, figure 13 a 85 μm pixel and calcification and figure 14 a 85 μm pixel and tumour tissue. These plots show that for the 100 μm system, the contrast enhancement improves with increasing magnification for calcification and tumour tissue except for the smallest radius detail considered. The situation is more complicated for the 85 μm system as the contrast offered by the CAXPCi system actually peaks for the 500 μm radius detail for system magnification values of between 1.25 and 1.29 depending on the fraction of illumination. This is because the effective source size increases with magnification. These simulations suggest that for the size of source likely to be encountered in practice, if the 85 μm pixel were employed, the magnification should not exceed 1.25.

5. Discussion

The final design of the system was chosen based upon the results of sections 2–4. The overall system length is the result of a trade-off between obtaining sufficient flux at the detector, space available in a clinical setting and the contrast improvement exhibited by the system. We settled on an overall system length of 2 m which we believe satisfies the aforementioned criteria, although we have also shown theoretically that a shorter length should be possible (Olivo and Speller 2007b). We decided upon the ANRAD a-Se (SMAM) flat panel detector, featuring a pixel size of 85 μm . As explained in section 2, a sufficiently low fraction of illumination cannot be obtained by a pixel of dimension 50 μm . This left the 85 μm and 100 μm pixels to choose from. The results of figure 10 show that the 85 μm pixel performs at least as well (generally better) as the 100 μm pixel when pixel spill-out is considered. Note that this plot is indicative of the results obtained with other imaging parameters. As we wish to maximize spatial resolution, the 85 μm pixel was chosen.

Having chosen the system length and pixel dimension, we next chose the source to sample distance z_{so} and thus z_{od} . The plots in figures 11-14 show that, particularly in the case of a source of 75 μm FWHM, M should not exceed 1.25 but, in all cases, the contrast improves up to $M = 1.25$. It is for this reason that $z_{\text{so}} = 1.6$ m and $z_{\text{od}} = 0.4$ m were chosen. It was decided to target a fraction of illumination as low as 0.2 in order to achieve the gains in contrast predicted by the simulation in section 4. With the already determined parameters, this lead to choosing values of $\eta_1 = 0.18$ and $\eta_2 = 0.27$ for aperture type 1 as shown in figure 4.

In practice, we had to allow a 4.5 cm gap between the detector and the detector mask. This is partially due to the sensitive part of the detector being within a housing and also to allow sufficient space to mount the detector aperture. Thus, the detector aperture dimensions were demagnified by a factor of 0.977 to accommodate this gap. Such a gap would not be required in a clinical device as the apertures would be incorporated within the detector. Small-scale tests apertures are currently being produced by Creatv MicroTech (www.creatvmicrotech.com) which will be used to verify the design experimentally before producing the large scale apertures.

6. Conclusions

We have presented the methodology for designing a CAXPCi system for application in mammography. The possible two-dimensional aperture designs were considered and 'L' type apertures were demonstrated to be optimal for the considered application. The aperture dimensions were then determined by considering x-ray flux and phase contrast sensitivity. Experimentally obtained data were used to model tumour tissue and calcifications. This model was used to optimize the system design which predicts that both will result in producing CAXPCi images of significantly higher contrast when compared with conventional mammography.

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Figure 1.

The system diagram of an CAXPCi system employing a point source. P is the detector pixel width and ΔP is the displacement of the detector aperture/detector arrangement relative to the projection of the sample apertures. The detector apertures are always centred upon the detector pixels. The object to be imaged would normally be placed adjacent to the sample apertures on the detector side.

Figure 2. Diagram illustrating the definition of the illuminated fraction in a one-dimensional case. In this example, the illuminated fraction would be a/b .

Figure 3.

Diagram demonstrating how the sample and detector apertures are described by a single cell as outlined in the figure. The detector aperture cell must have dimension matching that of the pixel (P) and thus the sample aperture must have dimension P/M .

Figure 4.

The top row of images show the unit cells of the three candidate aperture shapes with normalized dimensions. The sample and detector apertures are squares of side P/M and P respectively and so the physical dimensions are obtained by multiplying the normalized dimensions, η_1 and η_2 , by P/M and P respectively. The sample and detector apertures would be constructed by periodically repeating these unit cells. η_1 and η_2 are two fill factor parameters used to describe the apertures and e_1 labels an edge which is referenced later in the text. The lower row of images shows how an observer, positioned at the source, may see the effect of the combination of the sample and detector apertures if the detector apertures are shifted by amount $P\eta_1$ in both the vertical and horizontal directions. The opaque portion of the apertures have been depicted as being partially transmitting for illustrative purposes.

Figure 5. Intensity plots of σ_{cell} (left) and Φ_{cell} (right) for the permissible range of values of η_1 and η_2 for aperture 1. The region bounded by the solid line and indicated by arrows was identified for further investigation.

Figure 6. Intensity plots of the minimum fraction of illumination for a selection of values of W_{eff} , η_1 and η_2 for the three possible pixel dimensions.

Figure 7.

Plot of CAXPCi, conventional FSP and conventional absorption-based signals for a fibre of radius 0.1 mm composed of tumour tissue embedded in a 3.5 mm thick slab of 50% water and 50% lipid breast tissue. A pixel of 100 μm was employed with 100 scanning steps. The source had a FWHM of 50 μm , the illuminated fraction was 20% and the magnification was 1.25. If scanning were not employed, the acquired signal would depend on the position of the fibre with respect to the coded apertures. The vertical lines in the plots show the signals that would be acquired for one particular position of the fibre. Each signal has been normalized by the object free signal. The features enclosed in the broken boxes are due to photons interacting with a pixel not matched to the region of the sample aperture through which the photon entered the system.

Figure 8.

Plots for the same conditions as in figure 7 except that the lesion is composed of microcalcification tissue.

Figure 9.

Plots of the contrast improvement, relative to conventional absorption imaging, for calcifications (a) and tumour tissue (c) as a function of radius and relative to free space propagation imaging for calcifications (b) and tumour tissue (d). These results were obtained for a source FWHM of $50\text{ }\mu\text{m}$, $M = 1.25$, $P = 100\text{ }\mu\text{m}$ and a 50% water/50% lipid tissue slice. The interpolated lines are intended as visual guides only.

Figure 10.

Plots of the improvement in contrast relative to conventional absorption imaging for a tumour/calcification of radius $100\ \mu\text{m}$ assuming a 3.5 mm thick slice of tissue composed of a 50%/50% water/lipid combination. The results for the 33%/67% water/lipid combination were very similar and thus have not been plotted. These results correspond to a 20% illuminated fraction and $M = 1.25$, although the general trend shown in the plots does not depend on these parameters.

Figure 11.

Plots of contrast improvement relative to conventional absorption x-ray imaging against M for a variety of radii (R), fractions of illumination (η) and source FWHM for calcifications for a pixel dimension of $100\ \mu\text{m}$. The solid lines represent source FWHM of $25\ \mu\text{m}$, the dashed $50\ \mu\text{m}$ and the dotted $75\ \mu\text{m}$. Where a line does not extend over the entire range of M , it implies that the required fraction of illumination could not be achieved for the source FWHM.

Figure 12.

Plots of contrast improvement relative to conventional absorption x-ray imaging against M for a variety of radii (R), fractions of illumination (η) and source FWHM for tumour tissue for a pixel dimension of $100\ \mu\text{m}$. The solid lines represent source FWHM of $25\ \mu\text{m}$, the dashed $50\ \mu\text{m}$ and the dotted $75\ \mu\text{m}$. Where a line does not extend over the entire range of M , it implies that the required fraction of illumination could not be achieved for the source FWHM.

Figure 13.

Plots of contrast improvement relative to conventional absorption x-ray imaging against M for a variety of radii (R), fractions of illumination (η) and source FWHM for calcifications for a pixel dimension of $85 \mu\text{m}$. The solid lines represent source FWHM of $25 \mu\text{m}$, the dashed $50 \mu\text{m}$ and the dotted $75 \mu\text{m}$. Where a line does not extend over the entire range of M , it implies that the required fraction of illumination could not be achieved for the source FWHM.

Figure 14.

Plots of contrast improvement relative to conventional absorption x-ray imaging against M for a variety of radii (R), fractions of illumination (η) and source FWHM for tumour tissue for a pixel dimension of $85 \mu\text{m}$. The solid lines represent source FWHM of $25 \mu\text{m}$, the dashed $50 \mu\text{m}$ and the dotted $75 \mu\text{m}$. Where a line does not extend over the entire range of M , it implies that the required fraction of illumination could not be achieved for the source FWHM.

Table 1

Pixel spill-out for the C9732DK obtained from HAMMATSU and spill-out values for the SMAM obtained from Anrad.

	C9732DK (%)	SMAM-ANRAD (%)
Main pixel	100	100
First neighbour	49.0	13.0
Second neighbour	5.6	0
Third neighbour	0	0

Table 2

Raw and derived breast tissue data used to calculate the refractive index decrements δ and β at photon energy 17.5 keV. Data were obtained from three sources, *ICRU Report No 46* (ICRU 1991), Johns and Yaffe (1987) and Al-Bahri and Spyrou (1998). For the first three rows of data, μ/ρ , ρ and N_e are obtained directly from the source whilst in the fourth row, μ and N_e are obtained directly from the sources.

Tissue type	μ/ρ (m ² kg ⁻¹)	ρ (kg m ⁻³)	μ (m ⁻¹)	$N_e/10^{29}$ (m ⁻³)	$\beta/10^{-10}$	$\delta/10^{-7}$
Breast 50/50 (water/lipid)	8.64×10^{-2}	960	82.94	3.22	4.09	5.57
Breast 33/67 (water/lipid)	7.79×10^{-2}	940	73.23	3.15	3.61	5.45
Calcification	9.27×10^{-1}	3060	2835.09	9.18	139.96	15.87
Tumour			115.28	3.56	5.69	6.16

Table 3

Range of parameters considered in the simulations. z_{so} and z_{od} are the source to sample and sample to detector distances respectively as depicted in figure 1. W is the source FWHM, P is the pixel width, R is the tumour/calcification radius, η is the fill factor of the one-dimensional apertures, n_{detail} is the refractive index of tumour tissue and calcifications and n_{slice} is the refractive index of the tissue assumed to contain the tumour/calcification.

Parameter	Range	Unit
z_{so}	{1.50, 1.55, 1.60, 1.65, 1.70}	m
$z_{so} + z_{od}$	2	m
W	{0, 25, 50, 75}	μm
R	{5, 10, 25, 50, 100, 500}	μm
P	{85, 100}	μm
η	0.5	Unitless
n_{detail}	$\{1-15.87 \times 10^{-7} + i139.96 \times 10^{-10},$ $1-6.16 \times 10^{-7} + i5.69 \times 10^{-10}\}$	Unitless
n_{slice}	$\{1-5.57 \times 10^{-7} + i4.09 \times 10^{-10},$ $1-5.45 \times 10^{-7} + i3.61 \times 10^{-10}\}$	Unitless
Illuminated fraction	{0.1,0.2,0.3,0.4}	Unitless