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Can electronic zoom replace magnification in mammography? A comparative Monte Carlo study

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ABSTRACT. Magnification, which is considered to be a relatively high "dose cost" mammographic technique, is a complementary examination performed on women exhibiting breast complaints or abnormalities. Particular attention is given to the imaging procedure as the primary aim is to confirm the existence of suspected abnormalities, despite the additional dose. The introduction of post-processing capabilities and the widespread use of digital mammography promoted some controversy in the last decades on whether electronic zoom performed on the derived initial screening mammogram can effectively replace this technique. This study used Monte Carlo simulation methods to derive simulated screening mammograms produced under several exposure conditions, aiming to electronically magnify and compare them to the corresponding magnification mammograms. Comparison was based on quantitative measurements of image quality, namely contrast to noise ratio (CNR) and spatial resolution. Results demonstrated that CNR was higher for geometric magnification compared to the case of electronic zooming. The percentage difference was higher for lesions of smaller radius and achieved 29% for 0.10 mm details. Although spatial resolution is maintained high in the zoomed images, when investigating microcalcifications of 0.05 mm radius or less, only with geometric magnification can they be visualised.

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The carcinogenic risk associated with the delivery of high radiation doses such as those related to magnification views in mammography, in addition to the requirements for high image quality, have made it essential to optimise this technique. Although this radiation risk is considered to be relatively insignificant in the context of accurate diagnosis, work-up and treatment, an investigation has started for alternative techniques that could provide equivalent or improved characterisation of lesions and improved diagnostic information compared with that obtained from magnification views. The psychological ''cost'' of a woman being recalled for a second mammographic examination, the discomfort from the breast compression and the economic impact of an additional examination are also factors that have promoted research into alternative procedures that complement the information provided by standard mammography.

For many decades, magnification mammographic images of selected breast regions have been considered the most effective diagnostic tool for enhancing the visibility of subtle suspicious breast lesions and microcalcifications, thus providing improved diagnostic sensitivity and specificity. To this end, screen-film radiography was the gold standard for many decades and has now been replaced with digital radiography, which can also be combined with digital post-processing methods.

The enhancement of visibility in magnification views is attributed to the increase in contrast to noise ratio (CNR) caused by the increased fluence per irradiated area. The CNR increases with the degree of magnification, particularly for low degrees (increase of 75% between degrees 1.0 and 1.4) [5]. By contrast, a major disadvantage of magnification is the additional and significantly high dose of radiation delivered to the breast compared with the contact case. Owing to the fact that the breast is placed closer to the X-ray focal spot, both the entrance dose at the skin surface and the mean glandular dose (MGD) to the irradiated part of the breast are considerably higher than for the corresponding contact view. Typically, MGD is doubled at 1.5× magnification compared with a standard mammogram. Thus, there is an increased radiation risk [6, 7]. Regarding spatial resolution, this is significantly degraded as magnification increases owing to the finite dimensions of the focal spot and the detrimental penumbra effects [8, 9]. At the same time, however, spatial resolution is improved due to the effective detector resolution, which depends on the irradiated object's size on the detector plane [9]. For the low degrees of magnification usually applied in clinical practice, the

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overall system resolution is improved with magnification. However, for higher degrees it is degraded owing to the dominant effect of the focal spot dimensions [9, 10].

Among the new techniques introduced in the effort to replace magnification views, image post-processing, often facilitated by digital mammography, has become very popular [11–13]. Electronic magnification (zoom) of digital (or digitised) screening mammograms has recently come to the foreground of this research area and many authors are addressing this alternative. A question that arises is whether the image quality provided by electronic zoom is comparable to that provided by the (original) geometric magnification views. If not, another question arises – whether the dose-saving provided by electronic zooming can compensate for a potential detriment in image quality.

Several studies have been performed, most involving observers, to evaluate the image quality provided by the two techniques. Perisinakis et al [4] demonstrated that the enhancement of image features through post-processing (zooming) of both digitised contact images and geometric magnification mammograms equally improved the visualisation of subtle microcalcifications that are only rarely identified in standard full-field screen-film mammograms. Similar results have been reported by Vyborny et al [11], Smathers et al [14] and Powell et al [15]. These authors also showed that lesion visualisation achieved with geometric magnification mammograms (without the application of further post-processing) was similar to that achieved by electronic magnification and processing of the contact full-field image. Chan et al [16] showed that geometric magnification combined with stereotactic imaging in mammography provides better results than electronic display zooming of the contact stereotactic images.

Smith et al [17] included the radiologist's experience in their study; the authors demonstrated that, when evaluating microcalcifications, radiologists less experienced in mammography should not replace digitised and enhanced contact mammograms for microfocal-spot magnified mammograms. Other studies in this area have been reported in recent years [18–23] and, despite the fact that their conclusions vary, most exhibit a common characteristic: they are based on subjective human perception and decision criteria, known to vary significantly, rather than on objective metrics of image quality such as CNR and spatial resolution. Moreover, to our knowledge, to date no studies have been published comparing the primary image for both techniques without the application of additional post-processing methods (*e.g.* denoising or enhancement), based only on objective metrics of image quality.

In this study, a validated Monte Carlo model developed for producing simulated mammographic images under exposure conditions representative of clinical mammography was used. Sets of standard contact and geometrically magnified mammograms were produced using the same output. The contact mammograms were then electronically magnified (zoomed) and compared with the corresponding images produced with the geometric magnification with no further post-processing undertaken. The comparison was based on CNR (derived from signal and noise measured in the images and their background) and spatial resolution.

Methods and materials

Simulation of the procedure

The irradiation of the breast phantom and the test object was simulated with a validated Monte Carlo model, MASTOS [24–26]. This was developed by the authors for the purpose of deriving image characteristics in mammography.

The X-ray spectrum used was produced using a validated analytical model [27] with a 28 kVp tube voltage, a molybdenum target and a 0.030 mm thick molybdenum filter. The specific spectrum was found to provide the best overall performance for the simulation conditions considered, when compared with other anode/filter material combinations, and is representative of those used clinically for producing magnification images in mammography [5].

An automatic exposure control (AEC) system was simulated to mimic the mammographic clinical process and match the data from previous studies by the authors [5, 8]. The process consisted of adjusting the X-ray photon fluence, for each exposure condition considered, to deliver a 2 mR exposure at the image plane (for the CNR study) and a 5 mR exposure (for the spatial resolution study) for the field size considered and for both contact and magnification modes.

The exposure fluence map emerging from the phantom exit surface (pre-grid dose) was used to produce the spatially distributed digital signal and form the image (photons were binned for an area of $50 \times 50\ \mu m^2$ for the CNR study and for an area of $20 \times 20\ \mu m^2$ for the spatial resolution study). The simulated digital signals were mapped into an eight-bit grey-scale two-dimensional representation and displayed as a planar image.

The output signal exiting the phantom can be used as the input signal to a conventional screen-film or digital detector, such as those used for mammography. This results in an output signal from the complete imaging chain taking into account the detector performance (characteristic response and energy dependence). For this study, however, the imaging chain performance was not considered, as our aim was to investigate the two approaches for producing magnified images, their quantitative differences, potential and limitations.

To maintain computer processing time within normal values, the simulation considered only the region of the phantom that included the inhomogeneities (*i.e.* simulated lesions) for irradiation. This particular approximation has been validated in a previous study by the authors, both for contact and magnification geometries, and proved to have only minor effects on the resulting measured values [5]. Furthermore, this situation has similarities to the clinical magnification imaging procedure where only a selected region-of-interest of the breast is irradiated.

The Monte Carlo generated images were processed using dedicated home-made software (Medical Image Visualisation) developed for medical image post-processing and analysis applications [28].

Simulation of geometrically magnified images

Magnification in mammography is usually performed by positioning the breast closer to the X-ray tube (compared with a standard mammogram) at a focus-to-breast surface distance determined by the aimed degree of magnification. The focus-to-detector distance is maintained as in standard mammography and the antiscatter grid is removed. Detrimental effects of scattered radiation exiting the breast are reduced by the air gap between the breast and the detector [29]. An alternative geometry to provide magnification conditions involves increasing the source-to-detector distance compared with standard mammography; however, this technique is usually avoided. Although the whole breast volume is irradiated in standard contact mammography (full-field mammography), only the selected regions of the breast under investigation are imaged in the geometric magnification mode.

In the simulation studies performed, the focus-to-entrance ''breast'' distance was varied between 56 cm and approximately 26 cm resulting in magnification degrees between 1.0 (contact geometry) and 2.0, considering a 4 cm thick breast phantom. Various focal spot sizes (10 foci with areas ranging between 0.04 mm × 0.04 mm and 0.30 mm × 0.30 mm) and a single-peak Gaussian X-ray photon distribution were used to investigate the comparative influence of focus size on the spatial resolution of the images produced for each technique.

For the CNR studies, however, a broad focus of 0.30 mm × 0.30 mm was used in the contact mode and a fine focus of 0.10 mm × 0.10 mm was employed for the magnification views. The aim here was to mimic the characteristics of mammography systems available in clinical practice.

Electronically magnified images

Electronic magnification in mammography provides an enlarged view of the image; magnification is usually achieved by increasing the size of the matrix used to display the image data. In mammographic images, the image data corresponding to the region-of-interest under investigation in the contact image is displayed in a larger matrix. This results in a magnification factor determined by the ratio of the resulting and original image matrix sizes. Figure 1 illustrates the process of producing both the geometrically and electronically magnified images.

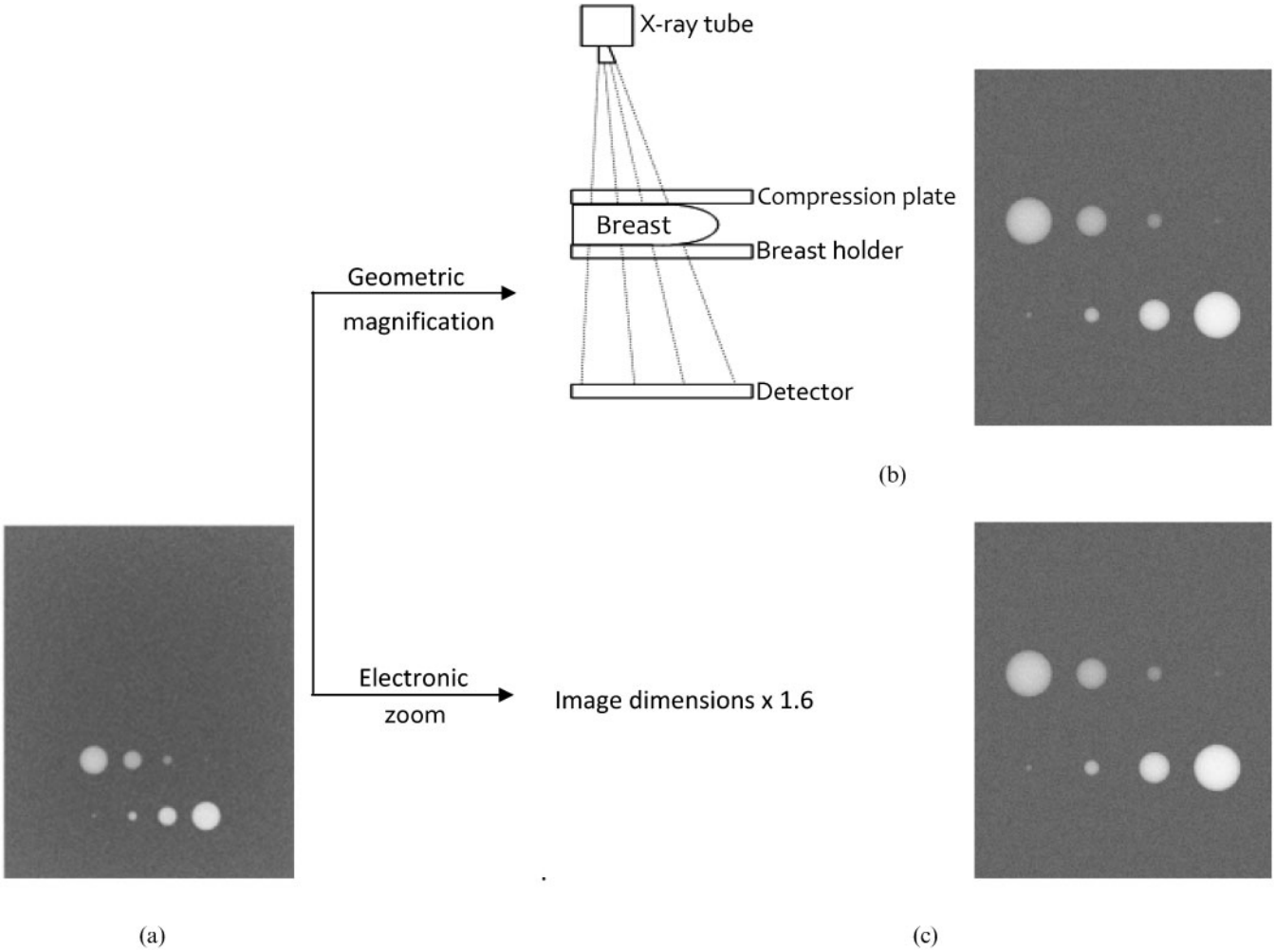

Figure 1. The process of producing geometrically and electronically (zoomed) magnified images of the mathematical breast phantom used in this study. (a) Contact geometry and contact phantom image derived using a spectrum produced with a 28 kVp tube voltage, a molybdenum target and molybdenum filter. (b) Geometric magnification and the same phantom image at 1.6 degrees of magnification. (c) The contact image derived from (a) zoomed at 1.6 degrees.

In the simulation study, the initial image matrix dimension (which was 1000 × 2000 pixels for images of the test object used for the calculation of spatial resolution and 1000 × 500 pixels for the breast phantom images used for the calculation of CNR) was multiplied by the desired zoom factor. This procedure was undertaken using a computer software package that provides image display and a post-processing zooming tool based on a linear interpolation method [30, 31]. Other methods (bicubic smoother and nearest neighbour) were also tested and showed no noticeable differences in the results.

Characteristics of the simulated test object and breast phantom

The breast phantom used for the CNR measurements comprised an homogeneous mixture of simulated adipose and glandular tissue. The relative composition of each tissue could be altered by varying the percentage glandularity between 0% and 100%, to simulate increased and decreased breast density.

The simulated phantom was semi-cylindrical in shape with a thickness and radius of 4 cm. The phantom contained two sets of five spherical inhomogeneities to mimic small-size high-contrast breast lesions (calcifications). Two different compositions for the inhomogeneities were considered — calcium oxalate (CO; $CaC_2O_4{\bullet}3(H_2O)$) and hydroxyapatite (HA; $Ca_{10}(PO_4)_6(OH)_2$) — and their radii varied between 0.05 mm and 0.75 mm. A drawing of the simulated phantom is shown in Figure 2.

Both sets of inhomogeneities were simulated in the centre of the phantom. Those composed of CO were placed at 0.7 cm from the simulated ''chest wall'', while inhomogeneities of HA were placed at 0.4 cm from it (Figure 2).

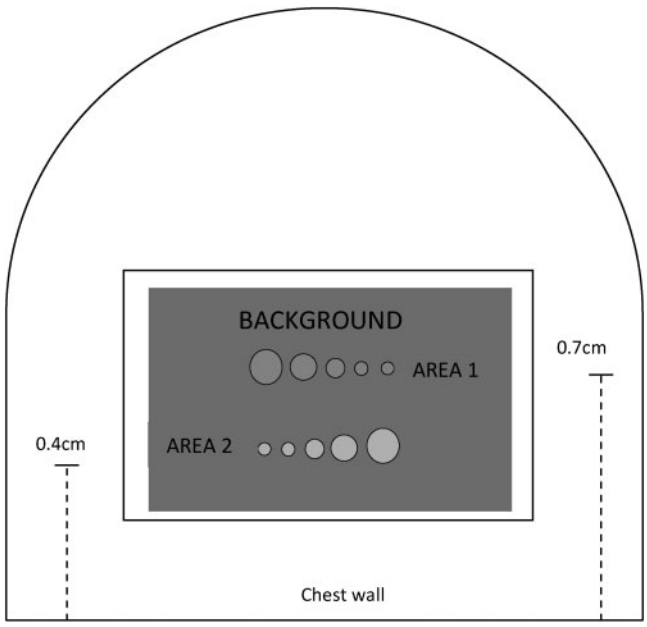

Figure 2. A drawing of the simulated phantom utilised for the contrast-to-noise ratio (CNR) studies. Inhomogeneities of area 1 consist of calcium oxalate set at 0.7 cm from the chest wall, while inhomogeneities of area 2 consist of hydroxyapatite and are set at 0.4 cm from the chest wall.

The test object used for the calculation of spatial resolution was a 4 cm thick sharp edge made of lead, in order to be absolutely non-transparent to X-rays. The object was positioned at the centre of the X-ray field with the central ray of the field vertical to the transition edge and the surface of the test object [8].

Calculation of image quality metrics

The CNR values for both geometric and electronic magnification modes were calculated using the formula proposed by Tapiovaara and Wagner [32]:

$$CNR = \frac{|C_{INS} - C_{BG}|}{\sqrt{N_{INS}{}^2 + N_{BG}{}^2}} \quad (1)$$

where C_{INS} is the mean grey-level value of the simulating insert, C_{BG} is the mean grey-level value of the background, N_{INS} is the noise measured in the simulated insert and N_{BG} is the noise measured in the background. To characterise the noise, the standard deviation of the grey-level value of the selected region-of-interest was considered. For the measurements of C_{BG} and N_{BG}, the irradiated part of the phantom around the inhomogeneities was used. Contrast was measured for each inhomogeneity within a pixel neighbourhood consisting of 15 × 15 pixels for the 750 μm inhomogeneity, 10 × 10 pixels for the 500 μm inhomogeneity, 5 × 5 pixels for the 250 μm inhomogeneity, 3 × 3 pixels for the 100 μm inhomogeneity and 2 × 2 pixels for the 50 μm inhomogeneity; the neighbourhood size increased respectively with the degree of magnification/zoom.

Regarding the calculation of spatial resolution, the edge method was used as described in a previous study [8]. The values reported in this study refer to the limiting spatial resolution (in lp mm^{-1}) corresponding to a threshold value (5%) for the modulation transfer function (MTF).

Results

In the following paragraphs, image quality metrics (CNR and spatial resolution) are calculated and compared for electronically and geometrically magnified mammograms.

Contrast-to-noise ratio studies

Using the aforementioned procedure, CNR was calculated from the geometrically and electronically magnified contact mammograms for a range of magnification/zoom degrees from 1.0 to 2.0 with 0.1 degree increments.

Figures 3 and 4 show the calculated CNR values for both magnification techniques and the various compositions and sizes of simulated lesions. In general, CNR associated with HA inhomogeneities resulted in higher values compared with those provided by the CO ones. For both techniques, CNR values increased with the geometric magnification/zoom factor. As discussed in a previous study [5], CNR increases with geometric

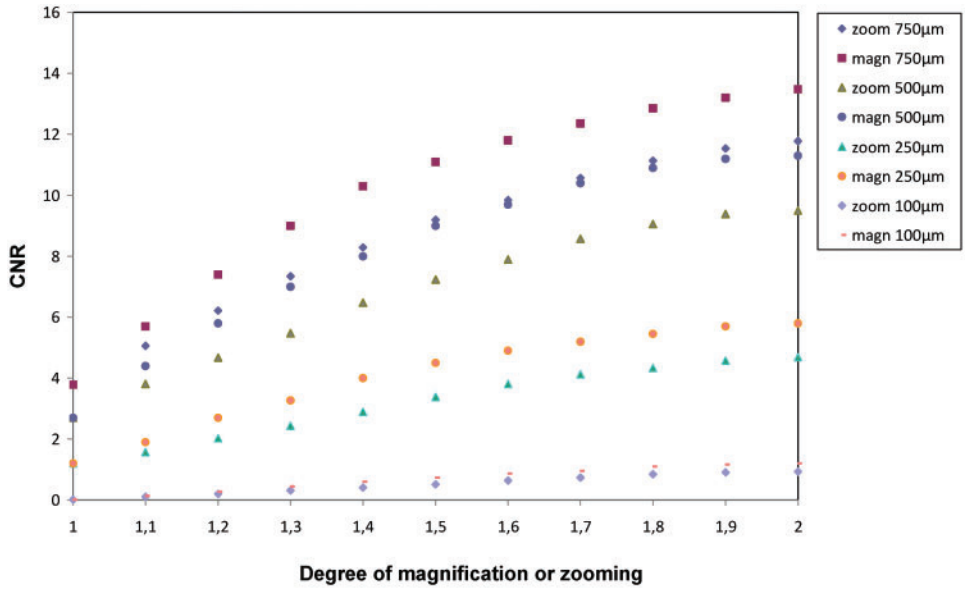

Figure 3. Contrast-to-noise ratio (CNR) values calculated for both electronic (zoom) and geometric (magn) magnification. The radii of the inhomogeneities range from 100 μm to 750 μm and consist of calcium oxalate.

magnification both because of an increase in image contrast and a decrease in noise (more photons contribute to the image formation). Contrast is also increased with the degree of magnification owing to the decrease in scattered radiation, as the breast is placed closer to the X-ray tube [5]. On the other hand, contrast did not vary with zoom, as expected, as there is no differentiation in the number of photons contributing to the image formation. However, there was a small decrease in the noise due to the linear interpolation performed by the zooming software and this resulted in an increase in the CNR.

This result confirms that electronic zooming of the contact mammogram does not provide additional object information to the initially acquired image, in contrast to the magnification views which do provide a significant increase in the subject contrast. CNR was 13% higher for geometric magnification as compared with the electronic method for CO inhomogeneities of 0.75 mm radius and factors of magnification between 1.0 and 2.0. The percentage difference increased for smaller radii and achieved 29% for 0.10 mm inhomogeneities. The corresponding percentage differences obtained for the HA inhomogeneities were 8% and 19%, respectively. Inhomogeneities of 0.05 mm radius are not presented in Figures 3 and 4, as their associated CNR values were below one – representative of a noise level higher than the corresponding signal. In fact, only with geometric magnification and factors higher than 1.7 could such small inhomogeneities be discriminated for both HA and CO compositions. The above finding clearly indicates that, as detail characterisation and discrimination conditions become more challenging (*e.g.* small radius inhomogeneities of CO), geometric magnification becomes more powerful. In the present simulation study,

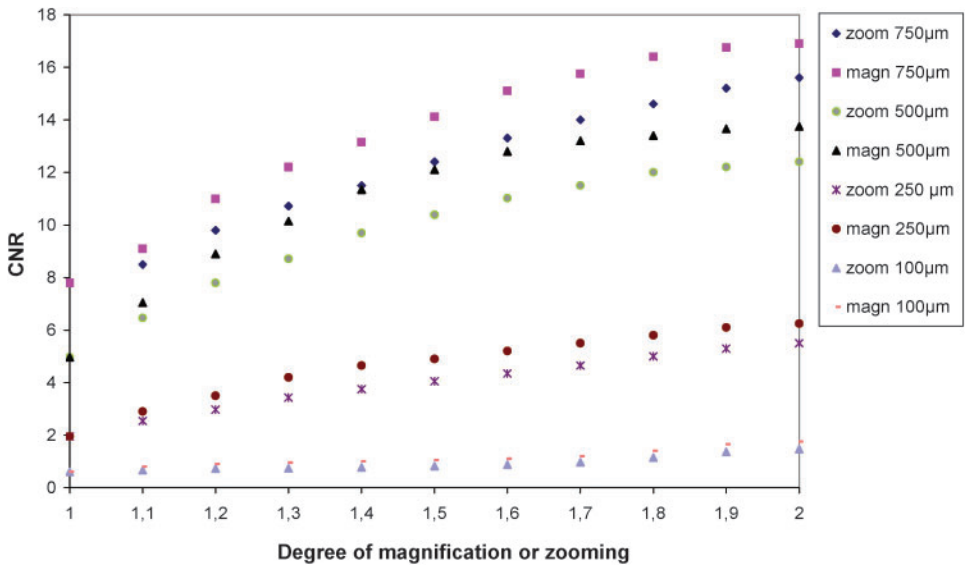

Figure 4. Contrast-to-noise ratio (CNR) values calculated for both electronic (zoom) and geometric (magn) magnification. The radii of the inhomogeneities range from 100 μm to 750 μm and consist of hydroxyapatite.

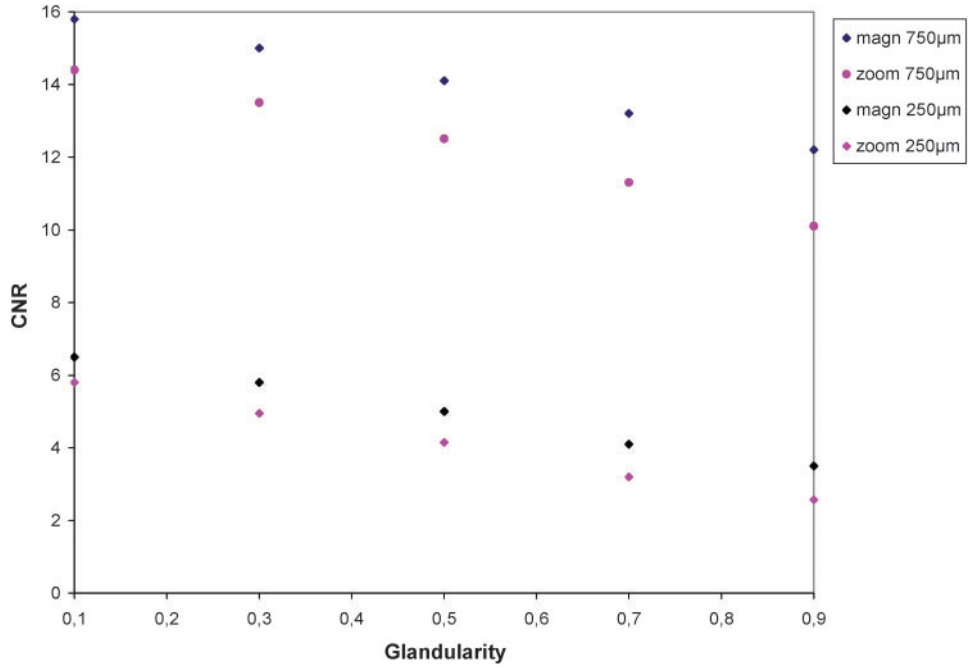

Figure 5. The effect of breast composition on the contrast-to-noise ratio (CNR) for hydroxyapatite inhomogeneities of two sizes (750 μm and 250 μm). Results are presented for both electronic (zoom) and geometric (magn) magnification. Calculations were performed using the molybdenum/molybdenum combination at 28 kVp, 1.6 degrees of magnification for zooming and a breast thickness of 4 cm.

inhomogeneities smaller than approximately 0.10 mm are visible with magnification, but are lost by zooming.

The two techniques investigated were also compared in terms of CNR for breast compositions varying between 10% and 90% in glandularity. Results are presented in Figure 5 for HA inhomogeneities of two sizes and show that CNR provided by geometric magnification is higher for all breast compositions investigated. The percentage difference found between the two techniques increased with the percentage of breast glandularity and is consistently higher for smaller sized inhomogeneities, as illustrated in Figure 6. This result confirms the aforementioned comment that, as discrimination conditions become more challenging (represented by simulating a denser breast), geometric magnification images provide improved CNR.

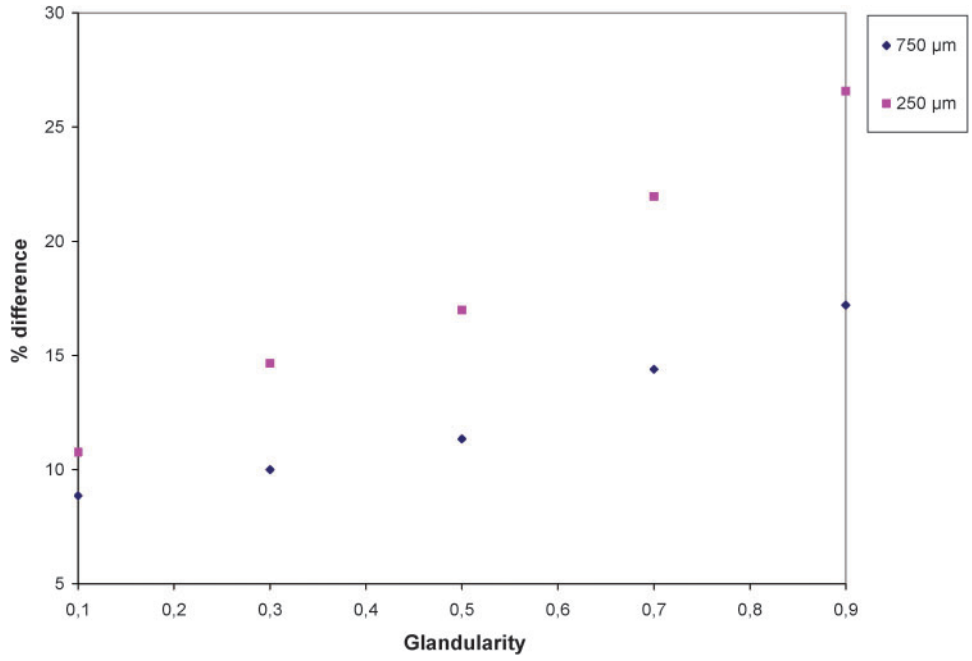

Figure 6. The percentage difference between zoom and magnification contrast-to-noise ratio (CNR) values for several breast compositions. The results presented are for hydroxyapatite inhomogeneities of 250 μm and 750 μm radii.

Spatial resolution studies

Various focal spot sizes (10 foci with areas ranging between 0.04 × 0.04 mm and 0.30 × 0.30 mm) and a single-peak Gaussian X-ray intensity distribution were considered in order to calculate the spatial resolution provided by the geometrically magnified and zoomed images, for degrees of magnification/zoom between 1.0 and 2.0. The spatial resolution values presented in this study represent the resolution caused mainly by the geometric blur, which represents a limitation on the spatial resolution, and do not take into account the effective resolution properties of the detector. As expected, spatial resolution is not affected when electronic zoom is applied and this is valid for all focal spot sizes considered. Zoom facilitates display such that there is a better visualisation of low frequency details. In contrast, geometric magnification contributes to a degradation of spatial resolution owing to the finite dimensions of the focal spot and the detrimental penumbral effects. An example of spatial resolution variations with magnification factor is illustrated in Figure 7, taking into account only the degradation caused by the focal spot size. The percentage difference between the spatial resolution values provided by the two techniques was 0, 36% and 52% for magnification factors of 1.0, 1.5 and 2.0, respectively. Such differences are crucial in mammography where a key aim is to detect micrometre-sized inhomogeneities obscured by clutter resulting from overlaying dense breast tissue. However, as previous studies have shown [9, 10], the overall system spatial resolution – considering both the geometric blur and the effective resolution of the detector – is increased for low degrees of magnification where the detector resolution impact is higher. The spatial resolution is decreased only at high degrees of magnification owing to the dominant detrimental effects of the focal spot finite size. Thus, by applying low degrees of magnification we can facilitate the visualisation of both low- and high-frequency details.

Discussion

The use of geometric magnification in mammography is a well established procedure, frequently used for further investigation of localised parts of the breast where suspicious abnormalities have been noticed in previous standard full-field mammograms. Despite the carcinogenic risk associated with the delivery of high doses, magnification views improve subtle lesion characterisation and diagnostic accuracy. Thus, this approach can only be replaced by techniques that provide improved or additional characterisation. Magnification results in increased CNR values. Spatial resolution is, on the one hand, degraded owing to the geometric blur yet, on the other, is improved due to the effective resolution of the detector. The net effect is positive and for low degrees of magnification the spatial resolution is improved [9]. Therefore, magnification is considered a valuable aid to contact mammography, despite the additional dose to the irradiated part of the breast.

As discussed in the introduction, several authors have reported no differentiation between the diagnostic accuracy provided by the geometrically magnified images and electronically magnified (zoomed) contact images. If this is the case, zooming offers the advantages that it would not contribute any additional radiation exposure to the patient and, in addition, could decrease the workflow and cost [4, 11, 14, 15]. These authors also showed that lesion visualisation provided by geometrically magnified mammograms (without further post-processing) was similar to

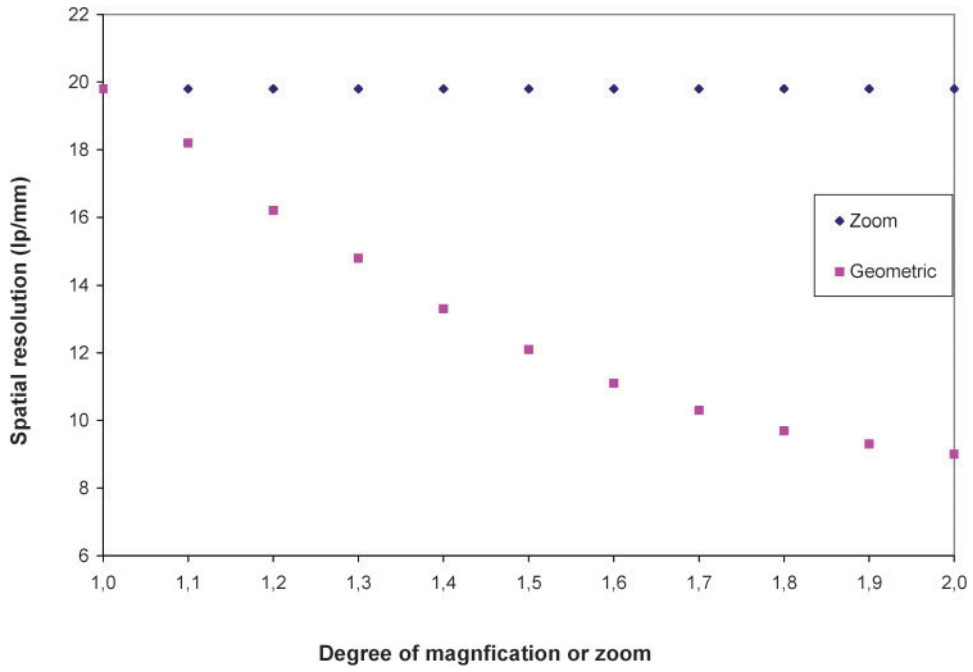

Figure 7. The influence of magnification and zoom on spatial resolution for a single-peak X-ray intensity distribution and for a 0.12 mm focal spot.

that provided by electronic zoom and post-processing of the zoomed contact image. However, most of these studies refer to low-frequency lesions and inhomogeneities, which usually do not represent a difficult task.

Other authors report a significant improvement in lesion detection when performing geometric magnification compared with electronic magnification of the contact images. Therefore, the use of the latter technique would introduce false-positives in the detection of calcifications [16, 17] (the latest of these publications is from Kim et al [19]). Irrespective of the varying conclusions of these studies, a common theme is that they are all based on human perception to evaluate and compare the potential of the two techniques. This introduces subjectivity into the results derived from varying human perception and decision criteria, although it is a human that will finally evaluate the image.

In this study, geometrically magnified mammograms were compared, in terms of CNR and spatial resolution metrics, with electronically magnified contact mammograms. In this way, the influence of human subjectivity was eliminated and raw exposure data were used to evaluate and compare the two magnification techniques without applying any further image post-processing. Observer performance and its relation to image-quality metrics is not discussed here, as this has been investigated in detail in previous studies. Moreover, the authors' aim here is to present absolute values of objective metrics for the two techniques.

Results showed that CNR was significantly higher in the geometrically magnified images compared with the electronically zoomed contact mammograms. The percentage difference between the two techniques was found to increase with the increased challenge posed by the detection conditions (*i.e.* when small radius inhomogeneities and/or dense breasts were considered). This result is representative of the significance of magnification mammography. Electronic zoom could be used to facilitate the visualisation of low-frequency abnormalities or those not hidden inside very dense breasts. However, in cases of small size and/or high-frequency lesions hidden in dense parenchyma, magnification can provide additional information that is crucial for lesion characterisation.

Additionally, when investigating the detection of a simulated microcalcification of 0.05 mm radius (or less), which was not visible in the contact mammogram, it was only possible to depict it with geometric magnification. As electronic zooming does not provide any additional information (no increase in subject contrast) there is no chance of visualising microcalcifications that are not visible in a full-field contact mammogram, although this is not really a task for magnification mammography. By contrast, because spatial resolution does not change with magnification it was found to be higher for the zoomed contact images than for those that were geometrically magnified, when only the geometric blurring effect was considered (not taking into account the effective resolution of the detector). According to this finding, a contact image could, theoretically, be magnified unlimitedly with zoom without resulting in any degradation of spatial resolution (and no improvement). However, even in this case, this capability would only be useful for large calcifications visible in the contact mammogram. In reality, the effective resolution of the detector is increased with the degree of magnification, owing to an increase in the apparent size of the irradiated object on the detector plane, and this effect is dominant for the low degrees of magnification usually applied in clinical practice [9, 10].

The post-processing of geometrically magnified breast images might also improve noticeably their diagnostic accuracy and is an area for future investigation. In addition to the clinical benefits of a particular magnification technique, other factors such as the discomfort and anxiety of the woman recalled when a magnification view is requested, impact on department workflow and economic cost are areas of interest to be investigated and objectively assessed.

This complete characterisation will provide a better understanding of the overall advantages of the magnification technique for mammography.

Conclusions

This study showed that geometrically magnified images can provide better image quality and, therefore, more accurate diagnosis of breast cancer than electronically zoomed images. This result was based on an objective investigation of raw data and image-quality metrics and is in accordance with published studies based on subjective human perception. Therefore, electronic magnification applied to full-field contact mammograms should not be used as an alternative to geometric magnification views. Electronic zoom is an excellent tool for the evaluation of an existing (screening) contact mammogram. However, under no circumstances can it replace the diagnostic value of magnification views because the zoomed image does not contain any additional information compared with the initial one.

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