Adapting Clinical Images to Appear with Different Noise and Sharpness to Model a Different Detector

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Abstract. Comparing the clinical performance of digital mammography technologies is challenging. The aim of this work is to develop and test a methodology for adjusting mammographic images taken on a given imaging system to simulate their appearance as if taken on a different system. Such methodology would be very useful for a wide range of system performance and design studies using both phantom and clinical images. The process involves changing the image blurring in accordance with the measured modulation transfer functions and adding noise (electronic, quantum and structure). The method has been tested by adapting flat field images acquired using an amorphous selenium detector and a computed radiography (CR) detector to different dose levels and comparing the resultant simulated NPSs with directly measured NPSs. For the detectors used in this work the NPSs at different dose levels are well predicted. This could be a powerful tool for studies of clinical image quality.

Keywords: mammography, simulation, noise.

1 Introduction

Extensive work has been undertaken on measuring the image quality of digital mammography systems. The physical parameters of images such as noise and sharpness have been measured using quantitative measurements such as modulation transfer function (MTF) and noise power spectra (NPS). Contrast detail measurements are used to ensure that clinical systems meet acceptable standards and have advantages in that they include factors in addition to the detector performance including the visual response of an observer, scatter, scatter rejection and beam quality [1]. Nonetheless, the relationship between contrast detail measurements and measurements of MTF and noise are well understood [2]. However, the relationship between the clinical task of cancer detection and the measured physical characteristics of the detector are less well understood.

Clinical evaluation of image quality in mammography is expensive and time-consuming. Clinical trials to compare the effectiveness of different systems are rarely conducted as they would require large numbers of patients to achieve both sufficient numbers of detected cancers and statistical significance. In particular it would be desirable to repeat exposures on the same breasts with the same compression to minimise confounding due to differences in breast appearance and compression but this raises ethical issues. Alternative methods involving some degree of image simulation have the potential to enable evaluations at reduced cost and time and without additional radiation exposure. For this purpose it is desirable to be able to acquire images on a given system and to simulate their appearance on a second system, so the performance of the two systems can be compared. This may be possible when the performance of the second system is inferior to that of the first system in terms of unsharpness and or noise, but not conversely. Such a method would enable the background tissue and compression to be matched in different arms of a study using the insertion of simulated cancers. Work has already been undertaken in correcting for dose and detector used [3-5].

The aim of this work therefore is to set up a framework for adapting images to appear with different imaging characteristics. In particular we are aiming to convert an image acquired using an amorphous selenium detector to appear as a generic CR image. Ultimately, after this has been shown to produce realistic test images, we would apply the methodology to a set of clinical images obtained with DR so that the performance of DR and CR systems for the detection of cancers can be compared at several dose levels.

2 Theory

2.1 Contributions of Electronic, Quantum and Structure Noise to the NPS

There are three different noise sources which contribute to the NPS: electronic, quantum and structure; each has a different relationship with dose and spatial frequency [6, 7]. Equation 1 shows the three components to the total noise power NPS_{tot} and their air kerma dependence. In this expression, NPS_e , NPS_q and NPS_s are the NPS components for the electronic, quantum and structure noise sources respectively at 1 μGy and K is the detector air kerma (DAK) in μGy .

$$NPS_{tot}(u, v) = NPS_e(u, v) + NPS_q(u, v)K + NPS_s(u, v)K^2 \quad (1)$$

2.2 Creation of Noise Images from Measured Noise Coefficients

With the knowledge of the NPS components shown in equation 1, a flat field image can be created. The first stage is to create three noise images for a DAK of 1 μGy based on the NPS components. The separate noise images are then combined (equation 3) to provide a simulation of the linearised flat field image without reference to an initial image.

$$I_M(x, y) = I_e(x, y) + I_q(x, y)\sqrt{K} + I_s(x, y)K + K \quad (3)$$

where I_M is the modelled image, I_e , I_q and I_s are the flat field images produced from the three noise sources, K is the DAK (μGy) (or the mean pixel value of the modelled image)

2.3 Adapting an Image to Simulate an Image Acquired with a Different Detector and Dose

The methodology is now extended to modify an image acquired with a given image detector with known MTF and noise power coefficients (from equation 1) to simulate images at different dose levels from a different detector.

The image adjustment was undertaken using the following steps:

- Change the original image sharpness to match the model image
- Simulate DAK change (if necessary)
- Adjust the noise components to match the system to be modelled

The original image is blurred by the ratio of the MTFs in frequency space. The three original noise sources are also blurred by the square of the ratio of the two MTFs to give NPS_b to account for the noise in the image being blurred.

Dose change: To simulate dose change in the detector the image is multiplied by a dose factor (R). The difference between the noise expected for the new dose level and the noise in the image which has been modified is shown in equations 5a and 5b. No adjustment is required for the structure noise as this noise source scales correctly with the dose correction.

Detector change: The next stage was to account for differences in noise between the two detectors. The noise differences ($NPS_{\Delta x}$) between the NPS of the three noise sources of the CR and DR systems are calculated (equation 4).

$$NPS_{\Delta x}(u, v) = NPS_{xm}(u, v) - NPS_{bx}(u, v) \quad (4)$$

$$NPS_{de}(u, v) = NPS_e(u, v)(1 - R^2) + NPS_{\Delta e}(u, v) \quad (5a)$$

$$NPS_{dq}(u, v) = NPS_q(u, v)(1 - R) + NPS_{\Delta q}(u, v) \quad (5b)$$

$$NPS_{ds}(u, v) = NPS_{\Delta s}(u, v) \quad (5c)$$

where $NPS_{\Delta x}$ is the differences between the detectors' NPS, NPS_{dx} is the total difference in noise between the original image and the image to be simulated.

The NPS_{dx} for the each of the noise source are made into an image using the method described earlier. Each pixel in the blurred image has been linearised to be equivalent to the detector air kerma, therefore an image with the correct magnitude of noise can be created by multiplying the blurred image with the noise on a pixel-by-pixel basis (equation 6). This will ensure that the noise is correct for that

dose and will have the correct frequency spectrum and the noise will still be correlated. Finally, the extra noise image (I_N) is added to the blurred original image to obtain the modelled image (equation 7).

$$I_N(x, y) = I_{de}(x, y) + I_{dq}(x, y)\sqrt{I_{blur}(x, y)} + I_{ds}(x, y)I_{blur}(x, y) \quad (6)$$

$$I_M(x, y) = I_{blur}(x, y) + I_N(x, y) \quad (7)$$

I_N is the noise array to be added to the blurred image, I_{blur} is the modelled image with the correct blurring but noise not corrected, I_M is the modelled image, I_R the dose reduced image, I_{blur} the blurred original image, and I_{de} , I_{dq} and I_{ds} the images produced from electronic and quantum noise sources and blurred by MTFs.

3 Method

3.1 Measurement of Imaging Characteristics of Digital Mammography Detectors

Two digital mammography imaging systems were used in this study:

- DR:** Hologic Selenia with an amorphous Selenium detector with pixel pitch of 70 μm .
- CR:** Carestream CR900 with GE Senographe DMR+ X-ray system with a CR detector with a pixel pitch of 50 μm .

For each system the measurements were made using methods as close as possible to those described by the IEC [8] using 2 mm high purity aluminium at the exit port of the tube, and radiographic factors of 28 kV, and a molybdenum/molybdenum target/filter combination.

The signal transfer function (STP) was measured by acquiring flat field images collimated to about 100 mm $\times$ 100 mm field over a range exposure values from a factor of five below to five above the typical dose (147 μGy for DR and 101 μGy for CR). The mean pixel value was measured for each image over 20 mm $\times$ 20 mm region of interest, 50 mm from the chest wall side of the image. The mean pixel value was plotted against DAK to give the STP relationship. All images were linearised with the inverse of the STP relationship. The resulting linearised pixel value then represented the DAK which facilitated image comparisons and manipulation. Using the same experimental set up as for the STP a further four images were acquired at normal dose plus quarter, half, double and quadruple typical dose level. The noise power spectra (NPS) were measured at each DAK level of the STP images.

The modulation transfer function (MTF) was measured using a steel edge 0.8 mm thick. The MTF was measured in the two orthogonal directions of the detector using IDL based 'OBJ_IQ_reduced' [9].

Using the NPS and STP images obtained over a range of dose, the separate contributions can be estimated. By fitting equation 1 to the NPS data obtained at the different dose levels, the three noise components are obtained as a function of spatial frequency for both systems. These values are in essence the NPS of the noise source at DAK of 1 μGy .

Using this knowledge it is then possible to predict the difference in the NPS for images taken under different conditions and to adjust the noise in the image on a pixel by pixel basis to model different imaging conditions.

3.2 Simulation of Flat Field Images at Different Dose Level from the Noise Coefficients

To test the image creation process, flat field images were modelled using the fitted noise components. The images produced using these methods were then compared with real images for the same factors. For this purpose, the normalised noise power spectra (NNPS) were calculated for each of the images and compared with the measurement from the original image.

3.3 Adjusting the Imaging Properties of an Image to Appear as if Acquired with a Different Detector

The flat field images used for measurement of NPS and STP were change an acquired image to appear as if acquired using a different detector.

Images acquired on the DR system at high dose were modified to appear as if acquired by the CR system. This was repeated for all five dose levels collected for measurement of the NPS. Images acquired on the DR system were then converted to images of similar dose acquired on the CR system.

A region of interest of 60 x 60 mm with a centre 50 mm from the chest was extracted from each of the DR images. The model images are then created using the method described in the theory section. The NNPS of the model images was measured and compared with those of the real images that were the model.

It was decided not to change the pixel pitch of the image in the conversion and so the modeled CR image had a pixel pitch of 70 μm instead of 50 μm . There is little to be gained in rescaling the image as there is no information in the original image above its Nyquist frequency. In reality the MTF of the CR system above Also the rescaling in itself may create errors [10].

4 Results and Discussion

4.1 Measurement of Signal Transfer Function and Modulation Transfer Function

The STP was measurements showed that the CR system has a logarithmic relationship and the DR system has a straight line relationship with an offset. With this information images from either system were linearised so that the linear pixel value will be equal to the DAK and so the images were comparable.

The MTF of the DR system is greater than the CR at all spatial frequencies. Thus one of the conditions allowing the images to be adapted to simulate another detector has been met for conversion from DR to CR.

Fig. 1. a) Signal transfer function of DR and CR systems as a function of pixel value against detector air kerma (K). b) The modulation transfer function for the CR and DR systems in the x and y axis.

4.2 Simulation of Flat Field Images at Different Dose Levels for a Given Detector

This was undertaken for five dose levels for both CR and DR. Fig. The DR images (fig. 2a) showed the NNPS of the real DR and the modeled image were closely matched over the whole dose range. The NNPS (fig 2b) of CR simulated images and real CR images agreement is slightly poorer than the modeled DR image results, but were still on average within a few percent between the real and modeled image.

Fig. 2. NNPS of flat field images created from measurements of the noises compared to the equivalent real image for a) DR and c) CR in scan direction

4.3 Adaptation of Image Noise to Simulate Image Obtained with a Different Detector

When converting from a high dose DR images (fig. 3a) to appear as CR images, then the NNPS of the modeled CR images closely matched the real CR image even down to low dose levels. It is easier to convert a large dose change in the image as the original noise in the image is reduced by the square of the dose change. Therefore, the

difference between the noise expected and in the dose reduced image is larger and so there is better scope to add the noise into an image with the correct magnitude and shape. However, when the dose change is small as in figure 3b, then there is less scope for accurately inserting noise. The largest difference between the real CR and the simulated CR image was at the conversion from a 28 μGy DR image to 26 μGy CR image. An issue arises at these dose levels as the DR system has more electronic noise than the CR system and at these dose level then electronic noise becomes a significant proportion of the overall noise in the image. The electronic noise cannot be simply removed and its effects are mitigated by a noise correction factor, without this correction the error would be much larger at this low dose level.

Fig. 3. a) NNPS of simulated CR image produced from DR image at 568 μGy compared to the real CR images; b) NNPS of simulated CR image produced from DR image compared to the real CR images, the dose change was 28 μGy to 26 μGy , 79 μGy to 75 μGy , 147 μGy to 145 μGy , 282 μGy to 210 μGy and 282 μGy to 300 μGy in x direction (CR scan direction)

The NNPS of the real CR images extends to higher spatial frequencies than the simulated CR image. The CR system has a Nyquist frequency higher than the simulated images as it was modelled from images obtained from a detector with a larger pixel pitch.

5 Conclusions

This work shows potential for the conversion of images both in terms of the detector used and the dose acquired. The work requires further improvement and validation. However, if this method is successful it will be a very useful tool for assessing the impact of using detectors with different imaging properties on cancer detection. Clinical trials currently require very large numbers of images to be able to detect differences in acquisition parameters, the reason for the large numbers is the large variability in clinical images due to differences between breasts being imaged. Even if the same breast is imaged using several technologies there will be differences due to the differences in compression/projection. This conversion program can allow the same image to appear as if it was imaged with a different detector and/or different dose. In addition it can remove variability due to differences in scatter rejection systems. This therefore may be a powerful tool for studies of clinical image quality.

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