Testing a wavelet based noise reduction method using computer-simulated mammograms

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ABSTRACT

A wavelet based method of noise reduction has been tested for mammography using computer-simulated images for which the truth is known exactly. This method is based on comparing two images at different scales, using a cross-correlation-function as a measure of similarity to define the image modifications in the wavelet domain. The computer-simulated images were calculated for noise-free primary radiation using a quasi-realistic voxel phantom. Two images corresponding to slightly different geometry were produced. Gaussian noise was added with a mean value of zero and a standard deviation equal to 0.25% to 10% of the actual pixel value to simulate quantum noise with a certain level. The added noise could be reduced by more than 70% using the proposed method without any noticeable corruption of the structures for 4% added noise. The results indicate that it is possible to save 50% dose in mammography by producing two images (each 25% of the dose for a standard mammogram). Additionally, a reduction or even a removal of the anatomical noise might be possible and therefore better detection rates of breast cancer in mammography might be achievable.

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INTRODUCTION

The purpose of this study was to test a newly developed method of reducing the contribution of non-correlated noise components to radiographic images [1]. The method is based on comparing two images obtained with slightly different geometry, but can also be used for two images sharing the same geometry. The images are compared at different scales, using a cross correlation function as a measure of similarity to define the necessary image modifications in the wavelet domain. Formally, a wavelet coefficient at some scale is given weighting proportional to the value of the corresponding correlation matrix in the given location [1]. This method is based on the idea of measuring anatomical noise [2, 3]. If the method is used on two images resulting from two detection processes of the same exposure, these should only deviate by means of quantum and detector noise. By using the proposed method on such images the quantum noise and the detector noise should be reduced greatly without reducing anatomical structures as long as these structures are not corrupted by noise. If two detection processes of two exposures with slightly different geometric conditions are the basis of this noise reduction, quantum noise, detector noise and anatomical noise are reduced by this method. Theoretically, the method could in this case disturb structures. It is the case which implies more difficulties to the noise reduction method based on correlations. Therefore, it has to be proven that there is no disturbance of the structures. This study tests the proposed method using computer simulated mammograms, for which the truth is known exactly. Also the noise reduction possibilities should be quantified. If the noise reduction method is even able to deal with this case of different imaging geometries it can be used for reducing technical noise but also anatomical noise.

MATERIAL AND METHODS

To quantify the effect of noise reduction and to prove the de-noising possibility a set of images with known noise components is needed. Such sets of images have been produced for this study by using computer-simulated images [4] of an artificial high-resolution voxel phantom of the breast (see figure 1) [5].

Due to the method of creation, this phantom is a very good representation of the structures and tissue compositions within the breast, and is used in this paper for simulation of the breast anatomical background.
Simulated images without quantum noise have been generated using Monte-Carlo simulation with a pixel size of 100 um assuming noise-free primary radiation.
Two such generated synthetic mammograms, corresponding to focus-detector distances of 650 mm and 600 mm, are shown in figures 2 and 3, respectively.
The synthetic mammograms do not contain highly realistic anatomical noise, due to the lack of fine, small-scale tissue structures in the phantom.
We added to each pixel Gaussian noise with a mean of zero and a variance per pixel proportional to the pixel value, which simulated the appearance of quantum noise.

Figure 1: A cross-section of the phantom, which have been used for the Monte-Carlo simulation.

Figure 2: Simulated mammogram using the phantom shown in figure 1 for a Monte-Carlo code using noise-free primary radiation with a Source focus detector distance (SDD) of 650 mm.

Figure 3: Simulated mammogram using the phantom shown in figure 1 for a Monte-Carlo code using noise-free primary radiation with a Source focus detector distance (SDD) of 600 mm.

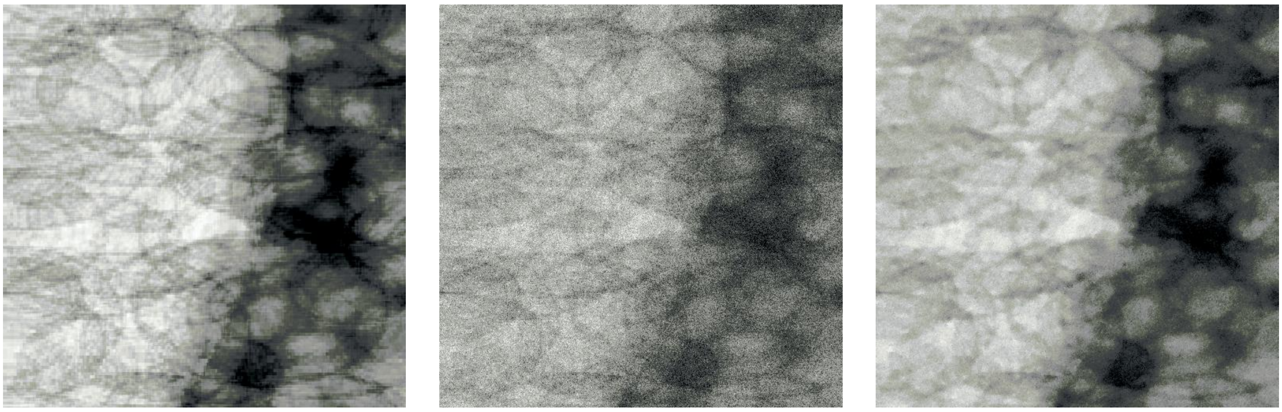
Gaussian noise was added to each pixel with a mean of zero and a variance per pixel proportional to the pixel value, which simulates the appearance of quantum noise. It should be mentioned here that the simulation is not referring to any detector noise or any frequency filtration caused by a possible detector as it could be described according to the model introduced by Cunningham [6].

A series of image pairs were obtained with noise levels in the glandular region of each image ranging from 0,25 % to 10 %. We evaluated for this study the four pairs with 1%, 2%, 4% and 10% additional noise to simulate something like a real mammogram (2% additional noise) and mammograms collected with much lower doses. The evaluation for the images with 1% added noise was selected for evaluation to illustrate the limitations in noise reduction and to give an impression of the maximal difference between the original image and the de-noised image. In this case, the noise in the difference image is expected not to be very prominent and thus better shows effects on the simulated anatomical structures than those with higher noise amounts. This assumption is correct as it can be seen in figure 9c versus figures 4d, 6d and 8c

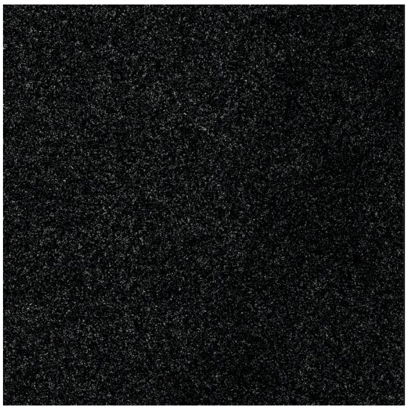
RESULTS

The effect of the de-noising as described in [1] is shown in the ROIs of images with 10% added noise (figure 4). In figure 5 the power spectra of the difference images (figure 4 D) are demonstrated compared to those of the added noise itself, which can be obtained by calculating the difference between images B and A.

The effect of de-noising on the appearance of the image is very evident. The noise can be reduced by 70% or more. This value results from taking the integrals under the curves of the power spectra of the added noise and the reduced noise. For this reduction a visual comparison of the re-constructed and the original images shows no noticeable corruption of the structures within the images. There might be only small effects of the noise reduction on the structures, which can also be seen on the appearance of the structures. However, it is very difficult to recognize the difference. It should be stated that 10% added noise is about a factor of 5 higher noise properties than what would be expected in real mammographic images. This corresponds to a dose reduction for a single image (two are needed for the image creation used as the de-noising process) of a factor of 25. We are not claiming to reduce dose in mammography by a factor like this, but it could according to these images be assumed that the relation between image quality and dose can be improved by a factor of 5 to 10 at least.



A B C



D

Figure 4: A: A part (ROI) of the original noise free simulated image; B: The same ROI with 10% added noise; C: The same ROI from the de-noised image; D: Difference image between C and A.

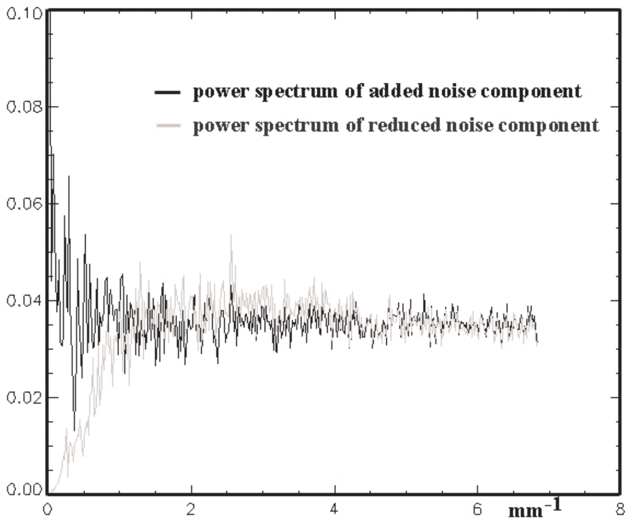
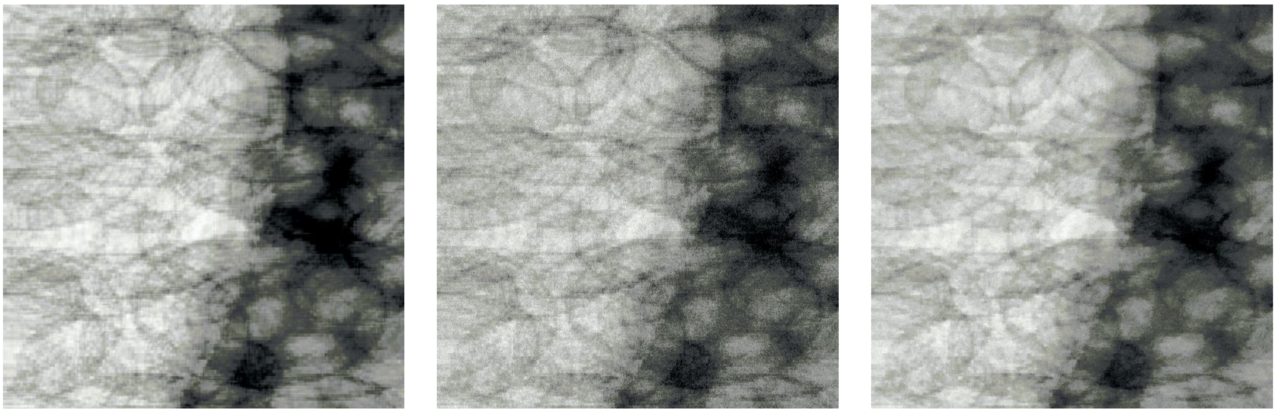


Figure 5**:** The power spectra of the difference images in figure 4 D compared to the noise spectra of the corresponding added noise.

Figure 6 shows the same ROI as analyzed in Figure 4 and its evaluation for the case with 4% added noise (Figure 7). Again the appearance of the image shows the effect of the de-noising very obvious. Hardly any corruption of the structures can be seen. The noise can again be removed by 70% or more if one is referring to a noise quantity for example like the integral under the determined power spectra as shown in figure 7. This figure shows also that the noise reduction is even more effective for higher frequencies. For the lowest frequency band the noise reduction has not been performed, instead the information from this band has been used to characterize the translation process needed for correlating the images in an optimal way. The assumption for this was, that the low frequency noise is not disturbing the detection of structures as much as the high frequency noise as can be concluded from results in a paper by Burgess and Judy [7] and as described in a paper by Hoeschen et al. [8].



A B C

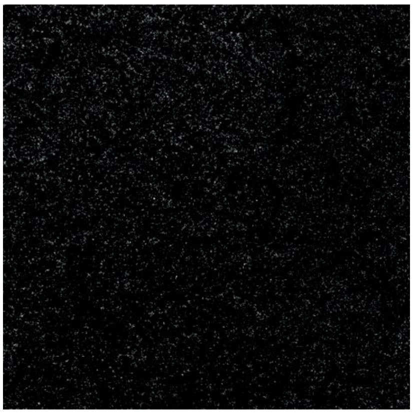


Figure 6: A: A part (ROI) of the original noise free simulated image; B: The same ROI with 4% added noise; C: The same ROI from the de-noised image; D: Difference image between C and A.

D

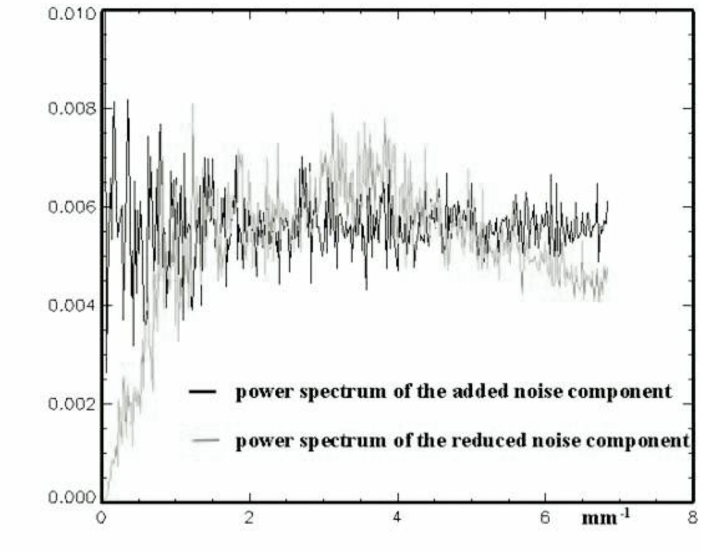
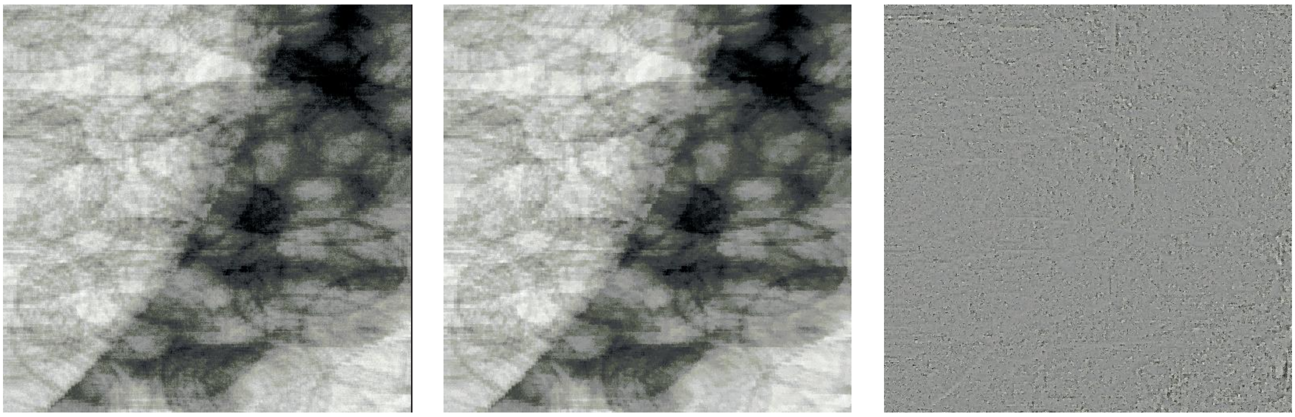


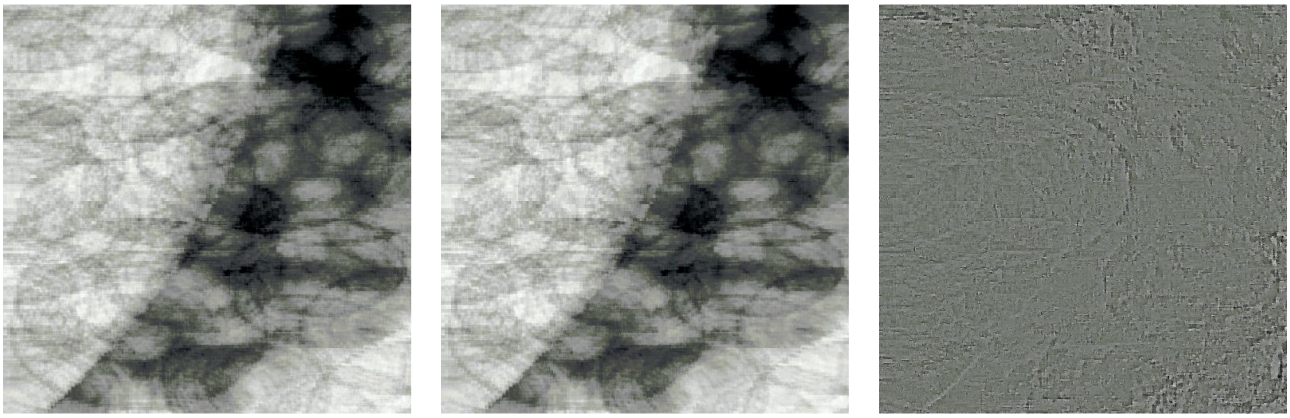
Figure 7: The power spectra of the difference images in figure 6 D compared to the noise spectra of the corresponding added noise.

Figure 8 and 9 show another ROI of the images for 2% and 1% added noise with the same impression. Noise is reduced, while structures are preserved. Figure part A for these two figures shows the image with the added noise while part B shows the de-noised image. As it has been expected for the images with 1% added noise the residuum (figure 9c) does show some edges of the structures as well. It can be clearly seen in the reconstructed image (figure 9b) that the structures themselves are preserved but the appearance of the edges is slightly stronger, which corresponds to a certain edge enhancement which is possible due to the noise reduction. This difference in the representation of the edges can be seen in the residuum image because the power of the noise is much lower than for higher noise levels, so that the differences in the edges have a visible contrast which is not the case for the residua calculated for higher noise levels.



A B C

Figure 8: A: A part (ROI) of the simulated image with 2% added noise; B: The same ROI from the de-noised image; C: Difference image (residuum) between B and A.



A B C

Figure 9: A: A part (ROI) of the simulated image with 1% added noise; B: The same ROI from the de-noised image; C: Difference image (residuum) between B and A.

CONCLUSIONS

The results of this study with images for which the truth is known exactly have shown that a recently proposed method of noise reduction [9] allows the reduction of the technical noise (quantum and detector noise) so effectively that it is possible to save at least 50% dose in mammography. This can be achieved by producing two images (each at 25% of dose). It might be possible to reduce dose even further but there seem to be theoretical limitations. These can be explained as follows:

According to data in the literature e.g. [10] one can assume that any detail in an image which should be detected and recognized by a human observer needs a signal to noise ratio of at least 3:1 for each single frequency component that should be analysed by the observer. There are data in the literature also, which would imply higher signal-to-noise-ratios needed for detection, however we used the lowest value, because this will give us a conservative expectation of the possible dose reduction. The method which is used for the de-noising process needs locally and frequency dependent a signal to noise ratio of 1:1 or higher in each of the images of the image pair to detect a structure without noticeable errors. When the signal to noise ratio gets smaller for a certain frequency, the algorithm used might detect a statistical correlation of the noise properties instead of the structures, which would be suppressed in this case. If the signal to noise ratio is larger than one this should not happen according to the process being used for the de-noising.

If one assumes that the noise is only caused by quantum noise, than a three times smaller signal to noise ratio as needed with the method giving the same image quality would correspond to a dose reduction by a factor of 9 for each single image of the two images needed, assuming a Poisson statistic for the quantum noise. However the detector noise is getting more important [11] for lower doses. This would mean that the amount of dose which can be saved is smaller. On the other hand it has been shown, that for some examinations noise effects due to overlaying structures in the patient which cannot be distinguished are of the same magnitude as the technical noise [3]. These effects are scaling with the dose which means they would be much smaller in the case of smaller doses and would allow an even further reduction.

The reduction of the technical noise by 70% might be suggesting a 3-fold increase of in the effective SNR, resulting again in a possible dose reduction of a factor of 9 for each single image in a two image procedure; however this value was gained using the whole frequency range. Assuming those frequencies with the most interesting structures in mammography and the biggest difficulties to detect, a much better noise reduction was achieved which implies that an even higher dose reduction would be possible.

The proposed noise reduction method [9] also allows the removal of this anatomical noise and thus has the potential to improve breast cancer detection rates for mammography as it has been stated in [12]. The reduction of anatomical noise could be proven by comparing the results presented here with simulated mammograms gained with a high resolution voxel phantom produced with using a real breast specimen. This voxel phantom is currently under development. [13]. It could be shown in this study that the structures are not significantly disturbed even when using different geometric imaging conditions for the two images.

On the other hand the method can be used to increase image quality as it has been shown on the 2% and 1% added noise images for example by using the implemented automated edge enhancement without suffering from increased high-frequency noise. This enhancement is again free of parameters and is automatically applied by the optimisation of the proposed method using the optimisation for the k-values as described in [9].

It should be stated here again very clearly, that by using this method the DQE of the detecting systems is not changed. The method just takes advantage of an additional information layer which has not been previously used. This information layer is the correlation between the relevant information content in both images.

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