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Enhancement of Breast Calcification Visualization and Detection Using a Modified PG Method in Cone Beam Breast CT

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Abstract

Cone Beam Breast CT is a promising diagnostic modality in breast imaging. Its isotropic 3D spatial resolution enhances the characterization of micro-calcifications in breasts that might not be easily distinguishable in mammography. However, due to dose level considerations, it is beneficial to further enhance the visualization of calcifications in Cone Beam Breast CT images that might be masked by noise. In this work, the Papoulis-Gerchberg method was modified and implemented in Cone Beam Breast CT images to improve the visualization and detectability of calcifications. First, the PG method was modified and applied to the projections acquired during the scanning process; its effects on the reconstructed images were analyzed by measuring the Modulation Transfer Function and the Noise Power Spectrum. Second, Cone Beam Breast CT images acquired at different dose levels were pre-processed using this technique to enhance the visualization of calcification. Finally, a computer-aided diagnostic algorithm was utilized to evaluate the efficacy of this method to improve calcification detectability. The results demonstrated that this technique can effectively improve image quality by improving the Modulation Transfer Function with a minor increase in noise level. Consequently, the visualization and detectability of calcifications were improved in Cone Beam Breast CT images. This technique was also proved to be useful in reducing the x-ray dose without degrading visualization and detectability of calcifications.

Keywords

Cone Beam Breast CT; PG Method; Breast Calcifications; Spatial Resolution

I. INTRODUCTION

Cancer has been one of the biggest threats to human life for many years and it is expected to become the leading cause of death over the next few decades [1]. In particular, breast cancer is one of the leading causes of death due to cancer among women [1]. Currently, there are no effective ways to prevent breast cancer because its etiology is still unknown. However, there are effective methods to cure breast cancer in early stages [2]. Therefore, early detection of breast cancer plays an important role in reducing its morbidity and mortality rates [2]. One way to detect breast cancer on its early stage is by studying and characterizing calcifications in the breast. Thus, it is important to utilize an image modality with a high 3D spatial resolution to better characterize the breast, in particular its calcification characteristics.

Cone Beam Breast Computed Tomography (CBBCT) is an effective and accurate diagnostic modality for breast cancer and calcifications. In particular, CBBCT has a sufficient high contrast spatial resolution to resolve calcifications as small as 200 μm which are one of the most important signs preceding breast cancer [3, 4]. Although their accurate characterization can usually be made on the basis of its radiological features in CBBCT images, calcifications usually are of different micro-sizes, shapes, and they are located in a textured background. Consequently, it is not easy to distinguish them from background noise in CBBCT images when using low dose levels. Therefore, it is valuable to enhance their visualization for a better characterization and a straightforward detection.

A lot of work has been done by different groups in order to improve the image quality of CBCT. Li's group proposed their curve-filtered FDK (Feldkamp-Davis-Kress) reconstruction algorithm in order to improve the quality of their circular CBCT image [5]. Ali's group developed a spatial domain filtering method applied to projection data to reduce the noise level and improve the uniformity in their CBCT system [6]. In this work, we also process the projection data, however, using an image extrapolation method in order to improve the CBCT image quality.

Image interpolation and extrapolation have been proved to be useful tools for improving image quality [7]. There are several commonly used interpolation methods such as linear interpolation and cubic interpolation. The Papoulis-Gerchberg (PG) method is an effective interpolation and extrapolation method presented by Papoulis and Gerchberg. It is an iterative algorithm of signal extrapolation and has been demonstrated to be effective in image restoration like in super-resolution (SR) and image in-painting [8].

In this paper, we first modified the PG method and analyzed its effect on image quality by measuring the modulation transfer function (MTF) and the noise power spectrum (NPS). Second, we acquired Cone Beam Breast CT images at different dose levels and pre-processed them by using the proposed technique to enhance the visualization of calcifications. Third, a computer-aided diagnostic algorithm, developed by Zhang [9], was utilized to evaluate the efficacy of this method to improve calcification detectability. Finally, we discussed the results of this study in the conclusion.

II. MATERIALS

For this study, a Flat Panel Detector (FDP)-based Cone Beam Breast CT imaging system, illustrated in Fig. 1, was utilized. Rad 71SP (Varian Medical Systems, Salt Lake City, Utah), the x-ray tube of this imaging system, has a 4" tungsten anode target and a focal spot size of 0.3 mm. Its pulsed output power is 9 kW at a 30% duty factor for up to 30 s and a voltage up to 49 kVp. For this study, the parameters were set to 49 kVp, 100 mA and 8.0 ms. Sedecal USA (Arlington Heights, Illinois) provides a modified SHF-class high frequency generator with 16-kW power (SHF-1635). It is fully programmable through an RS-232 link to preset x-ray technique parameters of kVp, mA, and pulse width time as short as 1 ms. It can also be triggered to synchronously pulse at 30 Hz.

This imaging system uses a PaxScan 4030 CB (Varian Medical Systems, Salt Lake City, Utah). This is a FPD that was specifically designed to meet the needs of cone beam x-ray imaging applications. The FPD consists of an amorphous silicon pixel matrix with a cesium iodide scintillator. The dual gain dynamic acquisition mode has a >16-bit dynamic range. Its active area is 397×298 mm² with a pixel pitch of 0.388 mm. The full panel can be operated at 30 frames per second (fps). 300 projection images were acquired over 360° at a rate of 30 fps using the dual dynamic 2×2 binning acquisition mode (388 μm pixel size). These projections were scattering-corrected by using the scattering correction method developed

by Cai [10]. Then, the 3D image data set was reconstructed using a FDK-based reconstruction algorithm with a linear-ramp filter. The reconstructed images have a voxel size of 155 μm . These settings were modified accordingly as noted below.

A breast CT performance/Quality Control phantom, illustrated in Fig. 2(a), was utilized. There are two inserts in this phantom (Fig. 2(b)). Insert 1 has a tungsten wire with a diameter of 50 μm and a tilted angle of 2° ; Insert 2 has six groups of calcifications (Fig. 2(c)). The tungsten wire was utilized for the evaluation of the high contrast spatial resolution in terms of the spatial resolution following the procedure of Betancourt [11]. The water area between Insert 1 and Insert 2 was used to evaluate the noise level as evaluated by Betancourt [12]. Insert 2 was utilized to evaluate the improvement of the visualization and detectability of breast calcifications using its calcification specks, whose sizes are 375 μm , 320 μm , 290 μm , 231 μm , 195 μm and 165 μm .

Also, a custom designed uncompressed breast phantom (CIRS, Norfolk, VA), illustrated in Fig. 3, was utilized. This phantom is composed of a background material equivalent to 50/50 fat/glandular tissue. It also has 10 embedded masses with sizes ranging from 1 mm to 10 mm in diameter with x-ray attenuation coefficient approximately the same as 100% glandular tissue. The breast phantom also contains embedded calcifications with sizes ranging from 0.21 mm to 0.35 mm. These calcifications are mainly clustered in two regions.

Finally, clinical data sets from one of our pilot studies were utilized. In particular, the data set illustrated in this study is primary important since it illustrates two calcifications clusters found in CBBCT from which, only one was located in their mammogram. In these images, as the calcifications are not on the same slice, the average of two adjacent slices was taken to demonstrate the results.

III. METHODS

A. Classical PG Method

It is well established that a finite object function has an analytic spectrum and a finite segment of an analytic spectrum determines the whole function in principle [13, 14]. As it is widely known, any signal has a unique Fourier spectrum. In other words, when given a segment of the spectrum of a finite object function, it is possible to recover the whole spectrum. This is the basic theory of the PG method.

The PG method is a useful tool that explores information beyond the Nyquist limit based on given portion of a signal or its spectrum. In reference [13], Papoulis presented a method for continuing the Fourier spectrum of band-limited function when given a segment of this function. He also proved the convergence of this method; after enough iterations, the truncated spectrum will be closer to the original spectrum. This method is simply an iteration which consists of four steps as illustrated in Fig.4. First, the Fourier Transform of the given segment $g(t)$ is taken yielding $G(w)$. Second, $G(w)$ is truncated by a window with a cutoff frequency of w to get $F_1(w)$:

$$F_1(w)=G(w)p_\sigma(w), p_\sigma(w)=\begin{cases} 1, & |w|<\sigma \\ 0, & |w|>\sigma \end{cases} \quad (1)$$

Third, the inverse Fourier Transform is applied to $F_1(w)$ and a signal, $f_1(t)$, closer to $f(t)$ is obtained. Finally, the segment of $[-T, T]$ is replaced with the known segment $g(t)$ and while keeping the rest of the signal unchanged:

$$g_1(t) = f_1(t) + [f(t) - f_1(t)]p_T(t), p_T(t) = \begin{cases} 1, & |t| < T \\ 0, & |t| > T \end{cases} \quad (2)$$

Then, step 1–4 are repeated until an accepted error is achieved. In reference [14], Gerchberg presented the same method in frequency domain. He further demonstrated the efficacy of this method on a two point problem with and without noise. The results indicated that this method reduces the error energy while achieving higher resolution.

B. Modified PG Method

In our CBBCT system, the images are acquired using the dual dynamic acquisition mode. In this mode, the FPD utilizes a low and a high gain switches and bins the data in a super pixel with a pitch of 0.388mm. It is assumed that this super pixel results from a 2×2 pixel block averaging of the projection image acquired through the high resolution (HR) acquisition mode of the FPD. In reference [8], the authors proved that the classical PG method will not deliver good results in this kind of situation. To solve this problem, they introduced a back projection method in order to insure that the super-resolved image obtained after every iteration conforms to the input low-resolution (LR) image. Based on their work, we modified the back projection algorithm so that it is suitable for our CBBCT system.

A 3-D CBBCT image is reconstructed from all the projection images acquired at different angles. However, the information of each projection image is limited by the detector Nyquist frequency. If more information could be explored for each projection image, the reconstructed image could obtain much more information. So that in our method, instead of processing CBBCT reconstructed images directly, the PG method was applied to the projection images in order to recover more information.

Another modification we made is that the LR projection image was enlarged to obtain an initial HR projection image using bicubic interpolation instead of setting the unknown pixel values to zero as in reference [8]. This can inhibit a large number of iterations needed making our method faster. Then, the 5-Step iterative computation method, illustrated in Fig. 5, was run: (1) Apply low-pass filter with a cutoff frequency σ to correct the HR projection image; (2) Down-sample the HR projection image to get its corresponding LR projection image; (3) Compute the difference between the corresponding LR projection image and the given LR projection image to obtain the error image; (4) Up-sample the error image to get HR error image; (5) Add the HR error image to HR projection image. The result of step 5 is labeled as corrected HR projection image. As more iterations are taken, the error becomes smaller and the corrected HR projection image gets closer to the expected projection image. The iteration is terminated until the error is smaller than a threshold. In our method, the iteration was terminated when the average pixel error is smaller than 1.5, because with more iterations the error decreasing rate drops significantly and the contrast to noise ratio does not improve effectively anymore.

The cut-off frequency σ in step 1 determines the highest frequency of the information that is expected to be restored. Different x-ray doses used in CBBCT system yield different amounts of information. Consequently, when processing projection images acquired using different x-ray doses, low-pass filters with different cut-off frequency σ must be selected for each case. In our method, we investigated different values for σ and selected the value that yields the best contrast to noise ratio.

Since the LR projection images of the FPD of our CBBCT system are binned through a 2×2 pixel block averaging from the HR projection image, Step 2 was implemented by averaging every 2×2 pixel block of HR projection image to form one pixel of its corresponding LR

projection image. In Step 4, the bicubic interpolation method was utilized. This method is more sophisticated compared to linear interpolation methods, because it tends to produce smoother edges than linear methods and suppress the noise in the error projection image.

IV. RESULTS

A. Image Quality

The breast CT performance/Quality Control phantom was utilized to quantitatively evaluate the image quality enhanced after pre-processing the projection images with the modified PG method of this study. Since one of the most relevant characteristics of image quality directly related to breast calcifications is the high contrast spatial resolution of the system, the MTF of our CBBCT system was evaluated following the same technique as Betancourt [11]. Fig. 6 illustrates the MTF of the CBBCT system before and after the modified PG method was implemented on the projection images. Clearly, the MTF was broadened with the modified PG method. In particular, the frequencies at 50%, 10% and 5% MTF increased from 0.916 lp/mm, 1.700 lp/mm and 1.914 lp/mm to 1.061 lp/mm, 1.905 lp/mm and 2.146 lp/mm, respectively. Consequently, after applying this method, the high contrast spatial resolution increased 15.83%, 12.06% and 12.12%, respectively which significantly enhanced the visualization of micro-calcifications. Similarly, it is possible to estimate the overall contrast enhancement by evaluating the ratio of the integral of the MTF. In this case, the ratio was $1.1747/0.9696=1.21$. So, there was an approximately 20% increase in overall contrast. Similarly, Fig. 7 illustrates the NPS of the CBBCT system acquired using the same technique as Betancourt [12]. In this case, the NPS were evaluated from reconstructed images acquired by using 80 mA with and without applying the modified PG method. The total noise power of the reconstructed images was 1.310 mm³ when the standard reconstruction method was used. On the other hand, the total noise power increased to 1.510 mm³ when the modified PF method was applied. This represented an increase of only ~15%. Overall, while the noise level increased by 15%, the contrast increased by 20%. Although our method increased the noise level, the percentage increase was less than the increase of spatial resolution. Consequently, image quality improved with respect to contrast to noise ratio. Since the reconstructed images will eventually be used for visual diagnosis, the visual improvement of modified PG method will be demonstrated in the following parts.

B. Cone Beam Breast CT Images

1) Phantom Study—Two phantoms with micro-calcified bodies mimicking breast calcifications were utilized in order to investigate the visualization enhancement of calcifications due to the modified PG method.

First, the specks in Insert 2 of the breast CT performance/Quality Control phantom were pre-processed using the modified PG method and the results are illustrated in Fig. 8. In Fig. 8, the phantom was scanned using 80 mA (a, b) and 125 mA (c) corresponding to an x-ray dose of 7.0 mGy and 10.6 mGy respectively. Fig. 8 clearly indicates the enhancement of the visualization of calcification after pre-processing the images with our modified PG method. For example, four sets of calcifications are visible in Fig. 8(a). However, the calcification sets in the rectangular area are incomplete as observed in Fig. 8(d). In this magnified image, it is evident that only three out of the six 0.231 mm calcifications are visible. In fact, for the 0.290 mm calcifications, it is difficult to distinguish the calcifications from background noise. On the other hand, after the modified PG method was applied, the calcifications were greatly enhanced as demonstrated in Fig. 8(b, e). In this case, all six 0.290 mm calcifications are enhanced; also, five of the 0.231 mm calcification are clearly visible. These results are comparable with those from Fig. 8(c, f) where 125 mA was utilized during the scanning process. In this case, although the noise of Fig. (a, d) is lower than that of Fig. 8(b, e), the

increased spatial resolution from the latter one enhances the visualization of calcifications. The contrast to noise ratio (CNR) was also measured in each case by calculating the contrast of the same calcification and the standard deviation of the background. The CNR measured from Fig. 8(a, b, c) is 12.51, 12.93 and 12.96 respectively. The plot of a line across the same calcification in each case is shown in Fig. 8(g), which indicates that modified PG method obviously enhanced the contrast of the calcification. Overall, using 80 mA with our modified PG method has a better much better visualization of calcifications than that of 80 mA without pre-processing the CBBCT images with the modified PG method. Furthermore, by comparing (e) and (f), the reconstruction with our modified PG method has indeed a similar calcification visualization. This means that our method has the potential to reduce x-ray dose by 33% without degrading the visualization of calcifications.

Second, the influence of our modified PG method was investigated by scanning a breast phantom using 80 mA. In particular, a mass with three calcifications was investigated as illustrated in Fig. 9. Although the modified PG method increased the noise level as illustrated in Fig 9(b, e), the contrast is enhanced making the boundaries of the internal structures sharper and their contrast higher than those from Fig. 9(a, e). In the magnified view of the mass in Fig. 9(c, d), it is evident that the visualization of calcifications is enhanced; the three calcifications at the mass boundary can be clearly visualized in Fig. 9(d) where as those in Fig. 9(c) are blurred and their contrast is lower. From Fig. 9(e), one can see that the modified PG method increased the contrast of calcification against soft tissues. Quantitatively, the CNR with and without modified PG method is 16.21 and 16.05 respectively. Thus, the modified PG method has the ability to enhance the image contrast and improve the visualization of calcification.

2) Clinical Data Study—After quantitatively and qualitatively analyzing the modified PG method for CBBCT images using phantoms, data sets from a clinical pilot study at the University of Rochester Medical Center were utilized to evaluate its efficiency with clinical data. One of such clinical data sets presented in this study, acquired at 49 kVp 160 mA, is illustrated in Fig. 10. In this particular example, the right breast has two clusters of calcification as diagnosed by a radiologist. This study indicates that, by using the modified PG method, CBBCT can better characterize the morphological aspects of both clusters. For example, in Fig. 10, the reconstructed images of clinical data and their magnified view of the calcifications with and without applying the modified PG method is illustrated. These images are illustrated in the same window level. Our modified PG method readily enhanced the calcifications in the clinical data since they appear much sharper in Fig. 10(b) than in Fig. 10(a). Also, their contrast from their surrounding tissues is increased. In the magnified views of these 4 calcifications in Fig 10(c, d), the enhancement can be more appreciated. In these two magnified views, those calcifications in Fig. 10(c) are blurred and their contrast is not as high as those from Fig. 10(d). Since the calcifications are sharper and their contrast is higher in the latter one, it is possible to better characterize the calcification with respect to their size which might yield a better understanding of their malignancy.

C. Improvement on Calcification Detectability

The Computer Aided Diagnostic (CAD) package developed by Zhang [9] was employed to detect the calcifications in three-dimensional CBBCT images. Briefly, this package utilizes thresholding combined with neural network techniques to segment and classify suspicious calcified bodies. This package has two main steps; first, a 3-D local thresholding plus a histogram thresholding were used to select suspicious calcifications from the background. Second, six features common to calcifications were extracted from each suspicious calcification and then they were fed into a trained Artificial Neural Network, which output a value between zero and one corresponding to the probability of being a calcification. Each

probability value was thresholded to judge whether or not the suspicious body is a real calcification. Accordingly, this CAD package was utilized to evaluate the effect of calcification detection after our modified PG method.

First, the specks in Insert 2 of the breast CT performance/ Quality Control phantom were fed into the CAD package to investigate the effect of the modified PG method on the detectability of the calcifications. Fig. 11 demonstrates the same slice as Fig. 8 of the detection results of the CAD package. Fig. 11(a) is the result of detection on QC phantom image without the modified PG method; while Fig. 11(b) is the result of detection on QC phantom image after applying the modified PG method. In this part of the study, two things are evident. First, the false positive outcome of the CAD package is decreased after implemented the modified PG method. For instance, in Fig. 11(a), the number of false positive is 38 with a sensitivity of 61.11% while that in Fig. 11(b) is only 8 with a sensitivity of 63.89%. This represents a False Positive decrease of around 80% with similar sensitivity. Furthermore, the CAD is unable to retain one calcification when the data was not pre-processed as illustrated by the missing calcification at the arrowhead in Fig. 11(a). However, after the application of the modified PG method, this calcification was retained; attesting its ability to improve the detectability of calcifications.

Second, the enhancement on the detectability of calcifications was analyzed by evaluating the Free-Receiver Operating Characteristic. This is a curve plotting the sensitivity versus False Positive (FP) number. This plot is generally used to report the performance of the detection algorithm [15]. A total of ten data sets, including five clinical data sets and five phantom data sets, were fed into our CAD system with and without the application of the modified PG method. While using the clinical data sets, only those calcifications marked by a radiologist were taken as true calcifications. Fig. 12 demonstrates the FROC curves of the CAD package with and without applying the modified PG processing to the data set. Clearly, the FROC with the application of the modified PG method lies above the FROC curve of standard CAD. This suggests that with the same number of False Positives, the CAD package has a higher sensitivity when using the pre-processed data with the modified PG method. In turn, this indicates that modified PG method effectively improves the performance of the CAD system. In this case, since the FROC curve is shifted upward by 8%, which is the ratio of the area between the two curves to the area under the standard FROC curve, it is possible to say that the calcification detectability increased by 8% if the modified PG is applied to the CBBCT data sets. Overall, the modified PG method has also the advantage of increasing the detectability of the calcifications by reducing the number of suspicious calcifications after Step 2 of the CAD package and increasing the value of the FROC curve.

V. CONCLUSION

In this study, the traditional PG method was modified to be implementable on CBBCT images. PG method is an iterative algorithm that continues the spectrum of a limit object to explore information. In the modified PG method, the bicubic interpolation method was utilized to initialize the HR projection images and then the iterative computation was performed. Each iteration of our method consists of 5 steps: (1) Apply low-pass filter to correct the HR projection image; (2) Down-sample HR projection image to get its corresponding LR projection image; (3) Compute the difference between the corresponding LR projection image and the given LR projection image to obtain the error image; (4) Up-sample the error image to get HR error image; (5) Add HR error image to HR projection image. In this way, the traditional PG method was modified according to the needs of CBBCT. After these modifications, the modified PG method enhanced the visualization and detectability of breast calcifications while reducing the dose level as well.

First, the modifications for PG method allow the implementation of this technique to CBBCT images. The difference between original projection image and the improved projection image was defined as the error image. Then, our method terminates the iteration when the average pixel error is smaller than 1.5. The initial bicubic interpolation allows for a quick pre-processing method since the results demonstrated that our implementation only uses 6 iterations for each projection image. Due to our modifications, the modified PG can be directly used on the projection images of our CBBCT system. As a result, our image quality is effectively improved by enhancing the calcifications.

Second, it was proven that this method increases image quality in CBBCT images. For example, the contrast of the system increases ~20% as the ratio of the integrals of the MTF before and after applying the modified PG method demonstrates. This 20% is larger than the increase of the total noise power increase that was only about 15%. Furthermore, the spatial resolution increase 15.83%, 12.06% and 12.12% as the ratio of the spatial frequencies at the given MTF values suggests. Due to this increase in image quality, the dose level could be reduced while keeping the visualization of calcification. This was demonstrated by Fig. 8(b, c) where the modified PG method was applied on a data set scanned using 80 mA and yields similar calcification visualization to that of 125 mA. Finally, our method increase the characterization not only of calcification but also of soft tissue as it was demonstrated by the sharpening and increase in contrast between the mass and background tissue of Fig. 9.

Third, the results indicated that our method improves calcification detectability. The increase of detectability is primary due to the enhancement of visualization of calcification since their contrast is significantly increased while the noise level is marginally increased. These differentiation of enhancement results in an 80% reduction of False Positives as Fig. 11 illustrates. Also, the True Positive of the CAD package increased as it was illustrated in Fig. 11(b) as well. In this case, this method can readily recover some calcification that could have otherwise been masked by the noise level. Finally, the FROC curve was evaluated for this CAD. It was demonstrated that the curve actually shifted at least an 8% of its current value. Thus, this method significantly enhanced the detectability of calcifications.

Finally, this method was applied to clinical data sets. It has been demonstrated that in these clinical cases, breast calcifications have been enhanced and the output of the CAD package was improved when applying the modified PG method. Overall, this method is significant in breast cancer diagnosis for two reasons; first, it has the ability to maintain the same image quality while reducing the dose level. This reduction of dose level, by itself, is a significant contribution since the dose level has recently been a hot topic among the medical community. Second, because this method increases the visualization of breast calcifications, radiologists can find and characterize them easier than before; and, Computer Aided Diagnosis packages, like the one developed by Zhang [9], will increase their sensitivity and their FROC. So, radiologists could have less suspicious calcifications to diagnose. In reference [16], Shaw's group presented the experimental results of their implementation of CBCT. Compared to their results, our system provides similar superior contrast between the three types of tissue: skin, adipose and fibroglandular tissue in true 3-D image for flexible viewing. However, using the same FP detector, we acquired images at 49 kVp instead of 80 kVp as in Shaw's group, and the voxel size is 155 μm after applying the modified PG method instead of 300 μm . The CBCT system of Shaw's group is able to detect calcifications as small as 280 μm at the dose level of 15 mGy. Compared to their system, our system does a better job by being capable of detecting 231 μm calcifications at the dose level of 7 mGy as discussed in part IV. B.

Overall, this method yields an optimal tool to help radiologist diagnose calcifications, either visually or with a CAD package, while reducing the dose level.

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Figure 1.
FDP-based Cone Beam Breast CT imaging system. It has a PaxScan 4030CB FPD and a Rad 71SP X-ray tube system.

Figure 2.

(a) The breast CT performance/Quality Control phantom. It is composed of two inserts. (b) Side view of QC Phantom. Insert 1 is used to measured linearity, low and high contrast resolution. Insert 2 contains the calcifications specks. (c) Cross-section of insert 1. Distribution of calcification specks.

Figure 3.
Custom designed uncompressed breast phantom.

Figure 4.

Sketch map of one iteration of Papoulis' method [13]. (a) Signal $f(t)$ that we want to recover in spatial domain. (b) Spectrum of $f(t)$. (c) Given segment of $f(t)$. (d) Fourier transform of the given segment $g(t)$. (e) Inverse Fourier transform of (f), (f) Low-pass filtered spectrum of $G(w)$. (g) Corrected signal of (e) using (c). (h) Spectrum of corrected signal in (g).

Figure 5.
Schematic diagram of computational procedure of Modified PG method.

Figure 6.
MTF of the CBBCT system.

Figure 7.
NPS of the CBBCT system.

Figure 8.

Reconstructed images of the breast CT performance/ Quality Control phantom. a) One slice of the reconstructed image using 80 mA without any image processing. b) The same slice of reconstructed image in (a) with our method. c) The same slice of reconstructed image using 125 mA without any image processing. d) Zoom-in view of the area marked in (a). e) Zoom-in view of the area marked in (b). f) Zoom-in view of the area marked in (c). g) Plots of the line across the same calcification in each case.

Figure 9.

Reconstructed images of the breast phantom. a) One slice of reconstructed image without improvement. b) The same slice of reconstructed image with the modified PG method. c) Magnified view of the the mass with three calcifications of (a). d) Magnified view of the the mass with three calcifications of (b). e) Plots of the line across the same calcification in each case.

Figure 10.

Clinical breast images of patient. a) One slice of reconstructed image without applying the modified PG method. b) The same slice of reconstructed image pre-processed with the modified PG method. c) Magnified view of the area marked in (a). d) Magnified view of the area marked in (b).

Figure 11.
A set of detection results the CAD detection package. a) Results without applying the modified PG method. b) Results after applying the modified PG method.

Figure 12.
FROC curves of CAD system with and without the modified PG method.