An investigation of a CT noise reduction using a modified of wiener filtering-edge detection

C Anam¹*, T Fujibuchi², T Toyoda³, N Sato², F Haryanto³, R Widita³, I Arif³ and G Dougherty⁴

¹ Department of Physics, Faculty of Mathematics and Natural Sciences, Diponegoro University, Jl. Prof. Soedarto SH, Tembalang, Semarang 50275, Indonesia.
² Department of Health Sciences, Faculty of Medical Sciences, Kyushu University, 3-1-1 Maidashi, Higashi-ku, Fukuoka 812-8582, Japan.
³ Department of Physics, Faculty of Mathematics and Natural Sciences, Bandung Institute of Technology, Ganesha 10, Bandung 40132, Indonesia.
⁴ Applied Physics and Medical Imaging, California State University Channel Islands, Camarillo, CA 93012, USA.
E-mail: anam@fisika.undip.ac.id

Abstract. The aims of this study were to investigate the noise reduction in a CT image using a modified Wiener filtering-edge detection method. We modified the noise reduction algorithm of a combination of the Wiener filter and edge detection by addition of a dilation stage after edge detection. We then evaluated kernel size of the Wiener filter, threshold values in the edge detection, and size of structuring elements in the dilation process. Images of adult anthropomorphic and self-built wire phantoms were acquired by the new 4-row multislice CT Toshiba Alexion™. The images of the anthropomorphic phantom were used for a visual evaluation, while the images of the wire-phantom were used to obtain the spatial resolution and noise of the images. A Wiener filter-edge detection filter coupled with dilation, potentially reduced more CT noise. We found that the spatial resolution and noise of the filtered images were influenced by the size of the Wiener filter kernel, threshold of edge detection, and size of structuring element.

1. Introduction
Several approaches have been proposed to reduce CT dose without compromising image quality. One method has been proposed is the tube current modulation (TCM) [1, 2]. In TCM, tube currents decrease and increase proportionally with the decreasing and increasing attenuation of body parts [3]. Tube current modulation could be implemented by the rotation of the x-ray tube (angle-modulation) or by modulation in the direction of the longitudinal axis (Z-modulation), or a combination of both [4]. Another method proposed for reducing the dose is to utilize iterative reconstruction (IR) [5], instead of filtered back-projection (FBP). In fact, the IR technique is not only iterative during reconstruction but also iteratively processes in either the sinogram [6] or image spaces [7], in accordance with the specific physical modeling or statistical approaches. There are several IR software products used by major CT vendors including ASIR, AIDR, VEO, IRIS, SAFIRE, and iDose [8]. However, the details of the algorithms are very sparse, and they are still considered proprietary algorithms [5].

Another method that can be used for CT dose reduction is the use of noise reduction in the image space [8]. A noisy image due to acquisition with a small tube current-time (mAs) parameter can have
its noise suppressed by a post-processing noise reduction algorithm, but this tends to reduce the spatial resolution of the image. Kalra et al. [9] reported that the application of noise reduction has the potential to reduce patient dose by approximately 50% with only a slight decrease in image quality. However, they did not explain the details of the six filters used. Other advanced noise reduction algorithms are available, including bilateral filtering [10], non-local means (NLM) filtering [11], and wavelet filtering [12]. Some of these methods even generate low noise images with a relatively good spatial resolution, but they require relatively heavy computation.

One proposal for a noise reduction algorithm that would maintain spatial resolution and use only light computation is to combine the Wiener filter with edge detection [13]. In this algorithm, the two processes are applied separately, and the resulting images are arranged selectively in order to choose low noise images with a relatively good spatial resolution [13]. This study aims to improve the method by adding dilation and evaluate the spatial resolution of these images with MTF curves.

2. Method

2.1. Post-Processing Filter

In this study, we modified the noise reduction algorithm that has been introduced previously [13], using a combination of the Wiener filter and edge detection. We added a dilation stage after edge detection so that the edge area would be wider compared to the use of edge detection only. Figure 1 shows the noise reduction algorithm. It was accomplished by two processes.

\[
h(x, y) = \mu + \frac{\sigma_L^2 - \sigma_g^2}{\sigma_L^2}(g(x, y) - \mu)
\]

where \(\sigma_g\) is the global variance of the noisy image, \(\sigma_L\) is the local variance of the image, and \(\mu\) is the local mean around each pixel. In homogeneous areas, \(\sigma_L\) is small, and the filter performs more noise...

---

**Figure 1.** The algorithm of the post-processing filtering using a combination of Wiener filtering and edge detection.

The first was the process of filtering the image using the Wiener filter, which is an adaptive post-processing filter which considers both the local and global variance of the image. The Wiener filter uses equation (1) to produce a less noisy image \(h(x, y)\) from the noisy image \(g(x, y)\).
reduction, and around the edges $\sigma_L$ is large, and the filter performs little noise reduction. The result is expected to be an image with low noise, without significantly decreasing the spatial resolution. In this study, the Wiener filter process was implemented pixel by pixel using kernels of 3x3, 5x5, and 7x7 pixels.

The second process was edge detection and dilation. In this research, the Prewitt method was used. The kernel of Prewitt was then convolved with the original image, $g(x,y)$, to find the gradient image in x-axis and y-axis. After obtaining the two-dimensional gradient image, a threshold value for the edges were determined. The pixel value was set equal to 1 for a gradient more than the threshold value which was considered the edge of an object, while below the threshold value the pixel value was set to 0 and was not considered an edge. From this process, we obtained the image depicting the edge of an object, $q(x, y)$. We evaluated the effect of varying the threshold value of edge detection, using values of 0.0001, 0.0005 and 0.0010.

From this edge detection process, boundaries between objects were obtained. A pixel value is 1 at boundaries among objects and otherwise is 0. However, the boundaries among objects may be only a narrow line. To widen the area of this edge, a dilation operation was used. The dilation process converts the pixel value of the image from 0 to 1 around the edge so that it becomes wider, $k(x, y)$. Dilation could be carried out using a structuring element (strel) with various geometries and various ranges. In this study, we used diamond-shaped strels with ranges of 2, 3 and 4 pixels. After the two separate processes were completed, a new image was composed using equation (2).

$$f = g \cdot k + (1-k) \cdot h$$

Equation (2) indicates that at the edge region or $k(x,y) = 1$, a new image is taken from the original image $g(x,y)$, while at the outside of edge or $k(x,y) = 0$, a new image is taken from the image that has been filtered with the Wiener filter $h(x,y)$.

2.2. CT Scanner and Phantoms

The evaluation of the modified the Wiener filter-edge detection was carried out using the images of phantoms scanned by the 4-row multislice CT Toshiba Alexion™ installed at the Department of Health Sciences, Faculty of Medical Sciences, Kyushu University, Japan. Two types of phantoms were used, the first was an adult anthropomorphic phantom and the second was a self-built wire phantom (Figure 2). The wire phantom comprised a CT injector syringe of 200 ml volume (Kyorindo Nemoto Co., LTD, Japan) filled with tap water, with a tin wire of diameter 0.1 mm along the center.

Figure 2. (a) Adult anthropomorphic phantom, (b) self-built wire phantom.

The anthropomorphic phantom was scanned in the head using axial mode. The images of the anthropomorphic phantom were used for a visual evaluation of the novel noise reduction method. The phantom was scanned with a tube voltage of 120 kVp, tube current of 100 mA, rotation time of 0.75 s, a field of view of 430 mm, slice thickness of 4x3 mm, and the image was reconstructed with a FC13 filter. The wire phantom was used to obtain the spatial resolution of the system using MTF curves [14]. The noise was calculated from homogeneous areas in this phantom image. The wire phantom was scanned along the center of the longitudinal axis in axial mode, using a tube voltage of 120 kVp, a field of view of 7 cm, rotation time of 1 s, and slice thickness of 4x4 mm. The wire phantom images were reconstructed using the FC13 filter (for soft tissue exams) and FC30 filter (for bone exams).

The regions of interest (ROI) for the MTF and noise calculations are presented in Figure 3. The center position of the ROI for the MTF calculation was at the center of the wire image, at position
(255, 290). The ROI size was 130x130 pixels. After obtaining the line spread function (LSF) in the x-axis, we found the MTF curve using the Fourier transform:

$$MTF(\omega) = |F(LSF(x))| = \left| \int_{-\infty}^{\infty} LSF(x) e^{-2\pi j \omega x} dx \right|$$  \hspace{1cm} (3)

where $\omega$ indicates the spatial frequency and $F$ indicates the Fourier transform.

The ROI for the noise calculation was 100x100 pixels, located in a homogeneous area of the image. The center of the ROI was at position (240, 210). The noise was calculated as the standard deviation of pixels in the ROI.

3. Results

3.1. Modified Wiener Filter-Edge Detection

Figure 4 shows the axial image of the anthropomorphic phantom (a) before filtering, (b) after filtering using the Wiener filter, and (c) after filtering using the modified Wiener filter-edge detection. The filtered image used a Wiener filter kernel of 3x3 pixels with an edge detection threshold of 0.0005, followed by dilation using a diamond-shaped kernel with a structural element (strel) range of 3 pixels. The spatial resolution of the image decreased significantly after Wiener filtering, with the edges of the temporal bones becoming noticeably blurred. In contrast, using the modified Wiener filter-edge detection, the spatial resolution remained as good as the original image, i.e., the edges of the temporal bones were still very sharp. The novel noise reduction generated images with low noise as if filtered by the Wiener filter alone.

Figures 4 shows the subjective appearance of the spatial resolution and noise in the images. A more objective description of the spatial resolution can be evaluated from the MTF curve, and noise can be calculated as the standard deviation in a homogeneous area. Figure 5 shows the MTF curves of the images of the wire phantom. The spatial resolution of the images filtered by the Wiener filter decreased significantly compared to the original image, whereas the spatial resolution of the images filtered by the modified Wiener filter-edge hardly changed at all compared to the original image. Values of MTF10%, MTF50%, and noise for each are presented in Table 1. The modified Wiener filter-edge detection produced MTF10% and MTF50% values that are higher than the Wiener filter alone. The modified Wiener filter-edge detection produced the same noise as the Wiener filter alone. The deviation standards of both were 2.17 HU and 6.15 HU for FC13 and FC30, respectively.
Figure 4. (a) Axial CT image of the head anthropomorphic phantom, (a) original image, (b) result of Wiener filtering, (c) result of modified Wiener filter-edge detection. The first row shows a zoomed view of the temporal bones (spatial resolution), and the second row is a zoomed view of the center of the images (noise).

Figure 5. MTF curves of the original image, after Wiener filtering and after filtering by the combination Wiener filter-edge detection.

Table 1. MTF10%, MTF50%, and noise of the original image, after Wiener filtering and after modified Wiener-filter-edge detection.

| Images      | FCI3      | FC30      |
|-------------|-----------|-----------|
|             | MTF10% (cycle/mm) | MTF50% (cycle/mm) | Noise (HU) | MTF10% (cycle/mm) | MTF50% (cycle/mm) | Noise (HU) |
| Original    | 0.86      | 0.47      | 2.63      | 1.12      | 0.86      | 9.07       |
| Wiener filter| 0.76      | 0.43      | 2.17      | 0.97      | 0.74      | 6.51       |
| Combination | 0.84      | 0.45      | 2.17      | 1.05      | 0.79      | 6.51       |

3.2. Evaluation of Modified Wiener Filter-Edge Detection

For a modified Wiener filter-edge detection, it is possible that spatial resolution and noise are influenced by many factors: kernel size of the Wiener filter, threshold values in the edge detection, and size of structuring elements in the dilation process. The impact of various kernel sizes of Wiener filter (3x3, 5x5, and 7x7 pixels) on the MTF curves are shown in Figure 6. The larger kernel size in the Wiener filter produced a lower spatial resolution in the image. Values MTF10%, MTF50%, and noise for each variation are presented in Table 2. It is clear that the larger kernel size also produces lower image noise.

The result of the modified Wiener filter-edge detection is not only influenced by the characteristics of the Wiener filter but is influenced by the threshold values used in the edge detection algorithm itself. In this study, we used a Prewitt mask with threshold values of 0.0001, 0.0005, and 0.0010. The
MTF curves for the various threshold values are shown in Figure 7, and the values of MTF10%, MTF50%, and noise are presented in Table 3. It is clear that increasing the threshold value reduced the spatial resolution of the image, but its effect on image noise is not straightforward. Dilation results in a wider edge around objects in the image, depending on the range of the structuring element (strel) used. In this study, we used strel values of 2, 3, and 4 pixels. The MTF curves obtained are shown in Figure 9. The values of MTF10%, MTF50%, and noise are shown in Table 4. Larger values of strel range produced a higher spatial resolution, although the effect is not as great as the effect of varying the edge detection threshold. Its effect on noise image is not straightforward.

Figure 6. The MTF curves for various kernel sizes of the Wiener filter (3x3, 5x5 and 7x7 pixels), (a) FC13 and (b) FC30.

Figure 7. The MTF curves for various threshold values in the edge detection algorithm (0.0001, 0.0005 and 0.0010), (a) FC13, and (b) FC30.

4. Discussion

The major limitation of the noise reduction is a poorer spatial resolution. To overcome this effect, several noise reduction methods have been developed which maintain the spatial resolution of the image [9-11]. A noise reduction technique combining the Wiener filter and edge detection has been reported [12], but not studied in detail. The spatial resolution of the image was conducted subjectively using a line-pair phantom. The study only used one value of kernel size in the Wiener filter and only one threshold value for edge detection [12]. The current study investigated the performance of the algorithm and improved the method by adding a dilation process after edge detection.
Figure 8. The MTF curves for dilation structuring elements of 2, 3 and 4 pixels, (a) FC13 and (b) FC30.

Table 2. MTF10%, MTF50%, and noise for various kernel size of Wiener filter for FC13 and FC30.

| Kernel Variation | FC13      | FC30      |         |         |         |         |         |         |
|------------------|-----------|-----------|---------|---------|---------|---------|---------|---------|
|                  | MTF10%    | MTF50%    | Noise   | MTF10%  | MTF50%  | Noise   | MTF10%  | MTF50%  |
|                  | (cycle/mm)| (cycle/mm)| (HU)    | (cycle/mm)| (cycle/mm)| (HU)    | (cycle/mm)| (cycle/mm)| (HU)    |
| 3x3              | 0.86      | 0.46      | 2.42    | 1.08    | 0.84    | 8.01    |
| 5x5              | 0.84      | 0.45      | 2.17    | 1.05    | 0.79    | 6.51    |
| 7x7              | 0.83      | 0.43      | 1.94    | 1.01    | 0.75    | 5.04    |

Table 3. MTF10%, MTF50%, and noise for various threshold value of edge detection.

| Threshold Variation | FC13      | FC30      |         |         |         |         |         |         |
|---------------------|-----------|-----------|---------|---------|---------|---------|---------|---------|
|                     | MTF10%    | MTF50%    | Noise   | MTF10%  | MTF50%  | Noise   | MTF10%  | MTF50%  |
|                     | (cycle/mm)| (cycle/mm)| (HU)    | (cycle/mm)| (cycle/mm)| (HU)    | (cycle/mm)| (cycle/mm)| (HU)    |
| 0.0001              | 0.86      | 0.46      | 2.17    | 1.08    | 0.84    | 8.63    |
| 0.0005              | 0.84      | 0.45      | 2.17    | 1.05    | 0.79    | 6.51    |
| 0.0010              | 0.83      | 0.43      | 2.17    | 1.01    | 0.75    | 6.51    |

Table 4. MTF10%, MTF50%, and noise for various sizes of structuring element.

| STREL variation | FC13      | FC30      |         |         |         |         |         |         |
|-----------------|-----------|-----------|---------|---------|---------|---------|---------|---------|
|                 | MTF10%    | MTF50%    | Noise   | MTF10%  | MTF50%  | Noise   | MTF10%  | MTF50%  |
|                 | (cycle/mm)| (cycle/mm)| (HU)    | (cycle/mm)| (cycle/mm)| (HU)    | (cycle/mm)| (cycle/mm)| (HU)    |
| STREL 2         | 0.83      | 0.44      | 2.17    | 1.04    | 0.76    | 6.51    |
| STREL 3         | 0.84      | 0.45      | 2.17    | 1.05    | 0.79    | 6.51    |
| STREL 4         | 0.85      | 0.45      | 2.17    | 1.07    | 0.82    | 6.51    |

Image spatial resolution and noise are influenced by many parameters in this method, including the kernel size in the Wiener filter, the edge detection threshold value, and the size of the structuring element used for dilation. An increase in the kernel size of Wiener filter results in a larger reduction of spatial resolution and noise. Since every pixel value is compared with a wider area of neighboring pixel values, it becomes more uniform. The more uniform pixel values lead to lower spatial resolution and noise. An increase in the threshold value during edge detection also causes a reduction in spatial resolution and noise. This is because fewer edges are detected so that the new image is taken more from the Wiener filtered image. An increase in the size of the structuring element causes an increase in spatial resolution and noise. This is because the edge area becomes wider so that the image is taken more from the original image. Optimization of these parameters is determined by the specific image characteristics, e.g., type of body part being imaged, and the type of reconstruction filters.
5. Conclusions
A combination of Wiener filter and edge detection, coupled with dilation, potentially reduced more CT noise. The spatial resolution and noise of the filtered images were influenced by the values of the size of the Wiener filter kernel, the threshold of edge detection, and size of the structuring element.

Acknowledgments
This work was funded by the Penelitian Dasar Unggulan Perguruan Tinggi (PDUPT) 2018, Ministry of Research Technology and Higher Education of the Republic of Indonesia, contract number: 532z/I1.C01/PL/2018.

References
[1] Gies M, Kalender WA, Wolf H, Suess C, Madsen MT. Dose reduction in CT by anatomically adapted tube current modulation. I. Simulation studies. Med Phys. 1999; 26: 2235-2247.
[2] Anam C, Haryanto F, Widita R, Arif I, Dougherty G, McLean D. Int J Radiat Res. 2018; 16: 289-297.
[3] Anam C, Haryanto F, Widita R, Arif I, Dougherty G. Information (Japan). 2017; 20: 377-382.
[4] McCollough CH, Bruesewitz MR, Kofler Jr JM. CT dose reduction and dose management tools: overview of available options. Radiographics. 2006; 26: 503-512.
[5] Beister M, Kolditz D, Kalender WA. Iterative reconstruction methods in X-ray CT. Phys Med. 2012; 28: 94-108.
[6] Hsieh J. Adaptive streak artifact reduction in computed tomography resulting from excessive x-ray photon noise. Med Phys. 1998; 25: 2139-2147.
[7] Winklehner A, Karlo C, Puippe G, Schmidt B, Flohr T, Goetti R, et al. Raw data-based iterative reconstruction in body CTA: evaluation of radiation dose saving potential. Eur Radiol. 2011; 21: 2521-2526.
[8] Kamezawa H, Arimura H, Shirieda K, Kameda N, Ohki M. Feasibility of patient dose reduction based on various noise suppression filters for cone-beam computed tomography in an image-guided patient positioning system. Phys Med Biol. 2016; 61: 3609–3636.
[9] Kalra MK, Maher MM, Sahani DV, Blake MA, Hahn PF, Avinash GB, et al. Low-dose CT of the abdomen: Evaluation of image improvement with use of noise reduction filters—pilot study. Radiology. 2003; 228: 251–256.
[10] Al-Hinnawi AR, Daear M, Huwaijah S. Assessment of bilateral filter on 1/2-dose chest-pelvis CT views. Radiol Phys Technol. 2013; 6: 385–398.
[11] Buades A, Coll B, Morel JM. A review of image denoising algorithms, with a new one. Multiscale Model Simul. 2005; 4: 490–530.
[12] Borsdorf A, Raupach R, Flohr T, Hornegger J. Wavelet-based noise reduction in CT images using correlation analysis. IEEE Transactions on Medical Imaging. 2008; 27: 1685-1703.
[13] Anam C, Haryanto F, Widita R, Arif I. AIP Conf Proc. 2015; 1677: 040004.
[14] Anam C, Fujibuchi T, Budi WS, Haryanto F, Dougherty. J Appl Clin Med Phys. 2018; 1-9. doi: 10.1002/acm2.12476.