



Application of a pixel-shifted linear interpolation technique for reducing the projection number in tomosynthesis imaging

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Abstract

Tomosynthesis images are reconstructed from several projections. However, the number of projections is proportional to the exposure dose, and a reduction in the number of projections would result in a reduction of radiation dose to the patient but also degradation of image quality. The purpose of this study was to propose a new computerized method to supply interpolation images instead of real projection images for maintaining the number of projection images and image quality of reconstructed tomosynthesis images. A set of images comprising one-half the number of projection images [37 projections (Half set)], selected from the original full set of projection images [73 projections (Full set)], was used at an interval of one by one. In this study, the authors used a new linear interpolation technique (Shift-Linear method), which takes into account shifted distances between two corresponding pixels on two projection images. The image quality of tomosynthesis images reconstructed from the full set and the quasi-full projection images, which were produced from the Half set using the Shift-Linear method, was compared. Image quality was assessed in terms of modulation transfer function, noise power spectrum, contrast-to-noise ratio, and the detective quantum efficiency. Using this proposed method, the image quality of reconstructed tomosynthesis images could be maintained with the reduction of approximately 50% exposure dose.

Keywords Tomosynthesis · Number of projections · Interpolation · MTF · NPS · CNR

1 Introduction

Tomosynthesis is a digital X-ray imaging technique that produces tomographic images by acquiring several projection images, at various angles, using several types of digital detectors, such as a flat-panel detector (FPD), and image intensifier. The term “tomosynthesis” has been used in reports since the 1970s; however, the acquisition techniques used in past and present tomosynthesis imaging differ considerably as a result of advances and improvements in X-ray detection systems (i.e., screen-film versus FPD) [1]. Digital

tomosynthesis reconstructs tomographic images from the acquired projection images using filtered back-projection and an iterative reconstruction method. The clinical advantages of tomosynthesis have been reported in many studies [2–4], which have demonstrated that this technique is capable of providing additional information along the depth direction in a two-dimensional X-ray image.

The advantages of tomosynthesis have been demonstrated in several studies that compared the technique with computed tomography (CT) in terms of the required patient radiation dose. For example, in a chest phantom study, the effective doses of conventional X-ray imaging and tomosynthesis were, respectively, 1.3% and 12% of that of CT [5]. Therefore, it is necessary to be aware that patient dose in tomosynthesis is ten times higher than that in general radiography.

To reduce patient dose during tomosynthesis imaging, several methods are potentially available. The most direct approach is to reduce the operating current when acquiring the projection images. This reduction in dose leads to an increase in image noise. Another approach to reduce patient dose is to reduce the number of projections

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for reconstructing tomosynthesis images. The number of projection images required for tomosynthesis imaging is directly related to the exposure dose and the image quality of the tomosynthesis images. Although a decrease in the number of projection images would result in a reduction of the patient dose, it increases image noise and degrades spatial resolution because of the lack of imaging data. However, we hypothesize that the number of projections can be reduced by replacing some of the projection images with interpolated images and, therefore, the loss of image quality of the tomosynthesis image reconstructed from those images can be minimized. We propose a new method to reduce the number of projections for reconstructing a tomosynthesis image by replacing some projection images with interpolation images aimed at reducing the number of projections. To our knowledge, there are currently no applications using interpolation images in the reconstruction of tomosynthesis images.

For image interpolation, in general imaging procedures, the conventional linear interpolation technique has been the most used basic image interpolation method [6–10]. However, in the specific case of tomosynthesis imaging, we assumed that the linear interpolation technique would cause image blur because of the difference in location between two corresponding pixels on two projection images. Therefore, we developed a modified pixel-shifted linear interpolation method that accounts for the shifted distance between the corresponding pixels located on two consecutive projection images (Shift-Linear method) for creating interpolation images. As discussed above, the conceptual idea underlying this method is to enable the use of interpolation images instead of real projection images and, thus, reduce exposure dose, with only slight quality degradation of the resulting tomosynthesis images.

2 Materials and methods

2.1 Materials

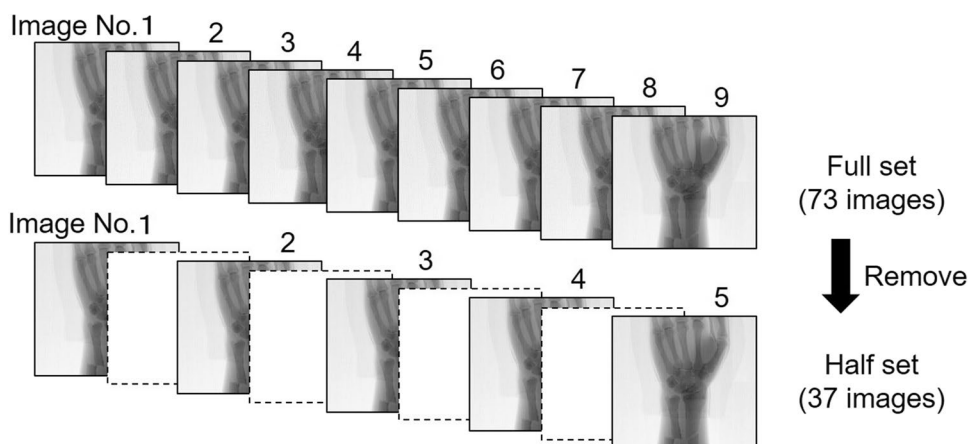
A digital tomosynthesis unit (SONIALVISION Safire17; Shimadzu Co., Kyoto, Japan) equipped with a direct conversion FPD, with a pixel pitch of 0.15 mm, was used. However, the pixel size of this apparatus is 0.30 mm due to the 2×2 binning used in the linear projections; therefore, a pixel size of 0.30 mm was used in this study. These projection images were obtained using a matrix size of 1024×1024 , with 12 bits per image. The X-ray tube had a real additional filter with 0.15 mm of Pb. To develop the platform for image reconstruction (based on the C language), the Visual Studio 2013 package (Microsoft Corporation, Redmond, WA, USA) was used. In addition, Radcal 9015 dosimeter with a 90×5 –6 chamber was used for dose measurement.

2.2 Image reconstruction

2.2.1 Image data set

The original image data set, consisting of 73 real projection images (Full set), was acquired using Safire17. The exposure conditions to acquire the Full set for each individual projection were 75 kV, 250 mA, and 2.0 ms. The integrated dose of all projections was used as the entrance dose at the corresponding height of the center of rotation of the X-ray tube. The entrance dose under these exposure conditions was 0.550 mSv. A data set comprising only one-half of the projection images (i.e., Half set [37 projection images]) was obtained by selecting alternating images (at one-image intervals) from the Full set (Fig. 1). It was assumed that the relative exposure dose for the Half set could be estimated to be 49% of the exposure dose in case of the Full set (the real entrance dose of 37 projections was 0.270 mSv at the center of rotation).

Fig. 1 A data set comprising half the number of projection images (37 projections) was selected from the Full set of projection images (73 projections) at an interval of one by one



To compare the degradation of image quality corresponding to the change in exposure dose, a full projection image data set with reduced (50%) exposure dose (Full50% set [73 projection images]) was also acquired.

2.2.2 Pixel value correction

For the projection images in tomosynthesis, the average pixel value for the same content was not consistent among all projection images. In this linear-type tomosynthesis, the focus-FPD distance for each projection changes due to X-ray tube movement in a direction parallel to the projection plane. Therefore, to estimate two corresponding pixels on two projection images for image interpolation, the pixel values of the original projection image had to be corrected using a pixel value correction curve (Fig. 2) determined in advance.

2.2.3 Method for creating an interpolation image

As mentioned above, the locations of two corresponding pixels in two consecutive projection images were slightly shifted along the travel direction of the X-ray tube because of X-ray tube movement. Therefore, for interpolating the projection images, it is necessary to account for the shifted distance between two corresponding pixels on two consecutive projection images in the travel direction of the X-ray tube.

The Shift-Linear method proposed herein is a modification of the conventional linear interpolation method. Although conventional linear interpolation uses the same pixel coordinates of two images for averaging pixel values, the Shift-Linear method uses the shifted distance estimated from two consecutive projection images for averaging two corresponding pixels. Figure 3a, b illustrate the pixel

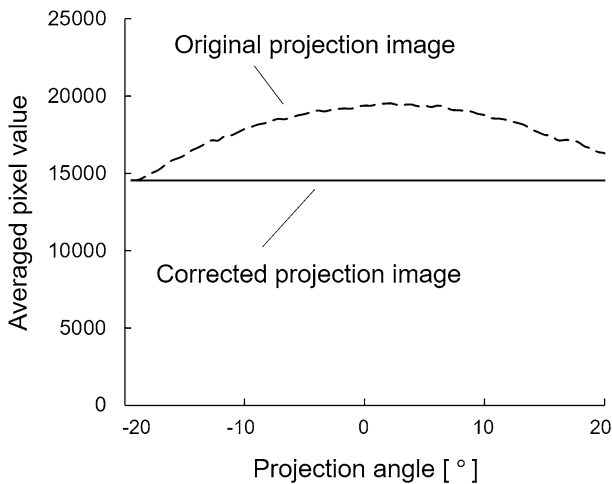


Fig. 2 Relationship between average pixel values in each original and corrected projection images among various projection angles

interpolation approach using the Shift-Linear method. First, the shifted distance between two corresponding pixels, P_A and P_B , was estimated using a template-matching technique. When applying template matching, corrected pixel values, described in the previous sub-section, were used.

In template matching using the corrected pixel values, only the Y direction (the travel direction of the X-ray tube) of the template scan was used. The kernel size of the template used in this study was empirically determined to be 17×17 . The location of pixel P_B was determined using the sum of the absolute differences (SAD) calculated with a template (P_A) [11]. Although the sum of the square difference (SSD) was also considered, this study used SAD for template matching because of its shorter calculation time compared with that of SSD [12]. The SAD was calculated using the following equation:

$$\text{SAD} = \sum_{x=0}^{X-1} \sum_{y=0}^{Y-1} |I(x, y) - T(x, y)|, \quad (1)$$

in which $I(x, y)$ and $T(x, y)$ are the pixel values at coordinates (x, y) of the corrected projection images A and B, respectively. The smallest SAD was used to determine the coordinate of pixel P_B on image B. The coordinates of the target pixel (P) on the interpolation image were determined using half the shifted distance. The value of pixel P was adopted using the averaged pixel value of P_A and P_B . The interpolated image at the target phase was created by applying this procedure to all pixels of the interpolated image.

2.2.4 Reconstructed tomosynthesis image

To evaluate image quality of the acquired tomosynthesis images, these images were reconstructed using the conventional back-projection (BP) method for the evaluation of resolution property, and filtered BP (FBP) method with the conventional Shepp–Logan filter kernel for the evaluation of noise property. Back-projection was used because the resolution property was strongly affected by the use of an FBP, whereas the real noise property could be measured only by the use of an FBP.

Five types of tomosynthesis images were reconstructed from the image set of the full 73 projection images (i.e., Full Set), two image sets of full 73 projection images, which consisted of a half number ($n = 37$) of projection images and their interpolation 36 images created using the Shift-Linear method (Quasi-Full set), and the conventional linear interpolation method (Conv-Full set), one image set of a half number ($n = 37$) of projection images (Half Set), and another image set of full 73 projection images with

Fig. 3 Illustration of the proposed interpolation technique (Shift-Linear method). **a** Location of two corresponding pixels, P_A and P_B , was shifted in the y direction due to the alignment of the X-ray tube and the detector in tomosynthesis. **b** Location and pixel value of a corresponding pixel on the interpolation image determined at half the distance of the shifted distance between two corresponding pixels, P_A and P_B , and their pixel values a_2 and a_2'

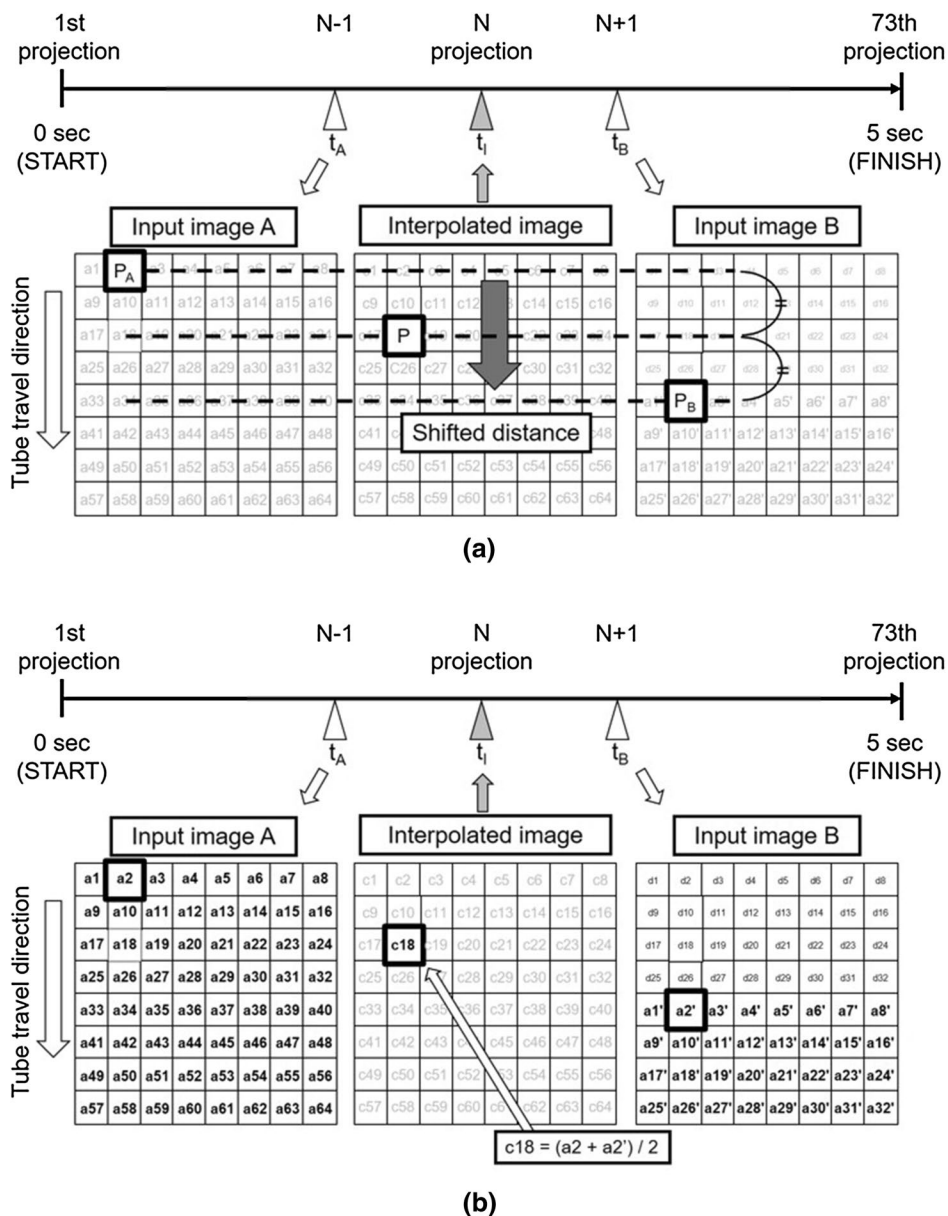


Table 1 Details of all data sets for reconstruction

	No. of real projection images	No. of interpolated projection images	No. of projection images used for one reconstruction	Relative total exposure dose for one reconstruction (%)
Full set	73	0	73	100.0
Quasi-Full set ^a	37	36	73	50.7
Conv-Full set ^b	37	36	73	50.7
Half set	37	0	37	50.7
Full50% set	73	0	73	50.0

^aInterpolation method used for this set is Shift-Linear method

^bInterpolation method used for this set is conventional linear interpolation method

50% dose of Full Set (Full50% Set) (Table 1). The Conv-Full set and Half Set were used to compare the image blur of the reconstructed tomosynthesis image with that of the Quasi-Full Set.

2.3 Assessment of image quality

The modulation transfer function (MTF), noise power spectral value (NPS), the detective quantum efficiency (DQE(u)), and contrast-to-noise ratio (CNR) of the images were used to compare the image quality of the reconstructed tomosynthesis images at the in-focus plane, with and without interpolation; these values were calculated as described below.

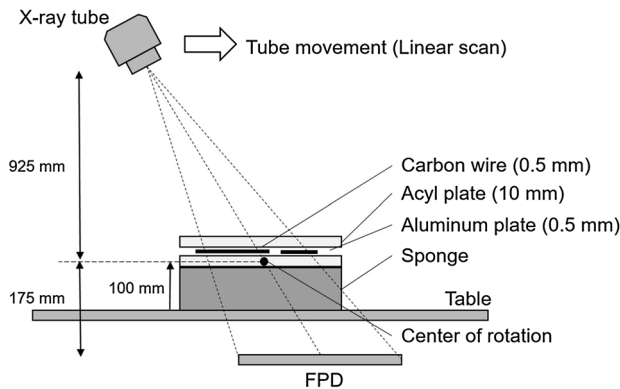
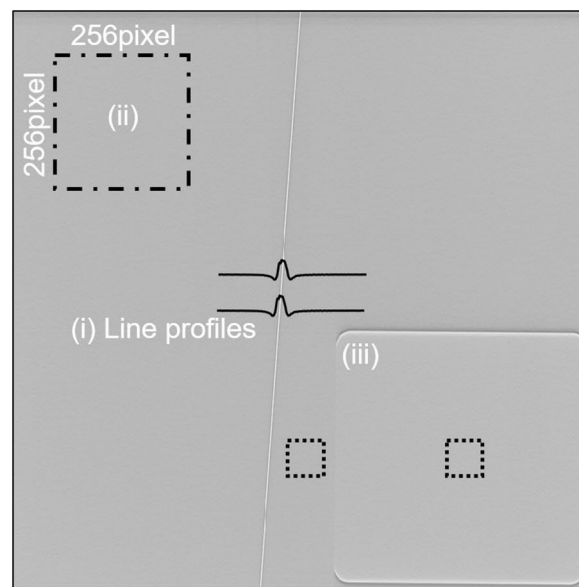
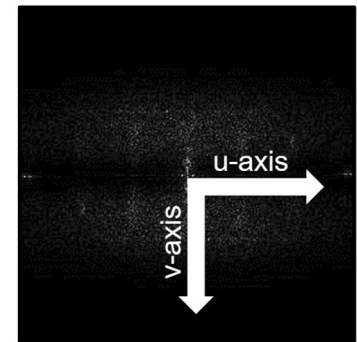


Fig. 4 Schematic of data acquisition of the projection image with the carbon wire, the aluminum plate and the acryl plates. The wire was slanted approximately 3° in the axis of the X-ray tube travel direction

Fig. 5 Reconstructed wire image of the tomosynthesis at the in-focus plane for calculating image quality. **a** The multiple line profiles (i) were used for the modulation transfer function (MTF). **b** The region of the direct X-ray (ii), which had 256×256 matrix size, was acquired and calculated using two-dimensional Fourier transformation for the noise power spectrum (NPS). The u - and v -axis were defined as shown in the figure **b**. The region of the aluminum plate ($100 \times 100 \times 0.5$ mm) was adopted for estimating the contrast-to-noise ratio (CNR) (iii). The region of interest was chosen with a matrix size of 50×50 for the CNR



(a)



(b)

2.3.1 MTF

The MTF was used to evaluate the degree of image blur in the tomosynthesis images caused by the lack of projection image information. The MTF was assessed using a thin carbon wire. This method was proposed in a previously published guideline for the evaluation of breast tomosynthesis [13]. The tomosynthesis image of the carbon wire (0.5 mm thick) used to assess the MTF used the acquisition geometry shown in Fig. 4. Acryl plates (10 mm) were used as a scatterer. The center height of the X-ray tube rotation was 100 mm above the table, which corresponds to the height of the acryl plate. The wire was placed parallel to the table, and the long axis of the wire was slanted to approximately 3° from the travel direction of the X-ray tube. An image of the wire is shown in Fig. 5; the acquired slice in this image corresponds to the height of the wire. To avoid the effect of a spatial filter (e.g., low-pass filter) in the MTF assessment, the BP image reconstruction method was used. As shown in the figure, to obtain the line profile with a small sampling interval, multiple line profiles were integrated (Fig. 5a). The MTFs of each reconstructed tomosynthesis image were obtained by applying a one-dimensional Fourier transform.

2.3.2 NPS

A two-dimensional fast Fourier transform method (2D-FFT), which is optimized by IEC (62220-1), was used to evaluate the NPS of the tomosynthesis image [14]. The NPS was also calculated using the tomosynthesis image of the wire at its in-focus plane, as shown in Fig. 5. The reconstructed tomosynthesis images used for NPS assessment were obtained using

the FPB method. The region of the flat field image, which consisted of a 256×256 matrix size without the wire image, was extracted (Fig. 5b). The NPS of the tomosynthesis image was estimated by removing an image trend from this image and applying the 2D-FFT.

2.3.3 DQE(u)

The DQE(u)s of each reconstructed image were estimated using the MTF and the NPS in the u -axis [15], as follows:

$$\text{DQE}(u) = \frac{\text{MTF}(u)^2}{q \cdot \text{NPS}(u)}, \quad (2)$$

in which q is the number of photons for X-ray beam quality used in the MTF and NPS measurements. In this study, to compare DQE(u) for all five data sets, reference numbers of photons were estimated for the four data sets (i.e., the Quasi-Full, Conv-Full, Half, and Full50% sets) based on the relationship between the exposure dose and the number of photons for the data set of Full set.

2.3.4 CNR

The CNR of the reconstructed tomosynthesis image was measured for each data set [16]. An aluminum plate ($100 \times 100 \times 0.5$ mm) was used for estimating the CNR. It was calculated using the aluminum plate image at its in-focus plane, as shown in Fig. 5. The CNR value is defined as follows:

$$\text{CNR} = \frac{I_{\text{signal}} - I_{\text{BG}}}{\sigma_{\text{BG}}}, \quad (3)$$

in which I_{signal} and I_{BG} are the average pixel values at the aluminum plate and image background, respectively. In addition, σ_{BG} is the standard deviation of the image background. The region of interest (ROI) located at the aluminum plate image region was chosen as a 50×50 matrix, as shown in Fig. 5. The average pixel value for I_{signal} was estimated using this ROI. The average pixel value and the standard deviation of the image background were calculated in a 50×50 matrix size at a distance far from the aluminum plate region.

3 Results

Results obtained for the MTF of the tomosynthesis images are shown in Fig. 6. The MTF of the Quasi-Full set was superior to those of the Conv-Full set, the Half set and the Full50% set, whereas it was slightly inferior to that of the Full set, which had 73 real projection images. When we focused on the spatial frequency corresponding to the MTF values of 50%, degradation in the MTFs of the Conv-Full (1.26 cycles/mm), Half (1.35) and Full50% (1.06) sets were clear compared with those of the Full (1.67) and the Quasi-Full (1.51) sets (Fig. 6b).

The NPSs along the u and v -axis are shown in Fig. 7. Compared with the NPS of a tomosynthesis image reconstructed from the Full set, that of the Full50% set was clearly inferior, whereas those of the Quasi-Full and Conv-Full sets were superior. The NPS of the Half set in the u -axis was slightly inferior to that of the Full set but was clearly inferior in the v -axis.

The results of the DQE(u)s estimated for each data set are shown in Fig. 8. The DQE(u) of the Quasi-Full set was

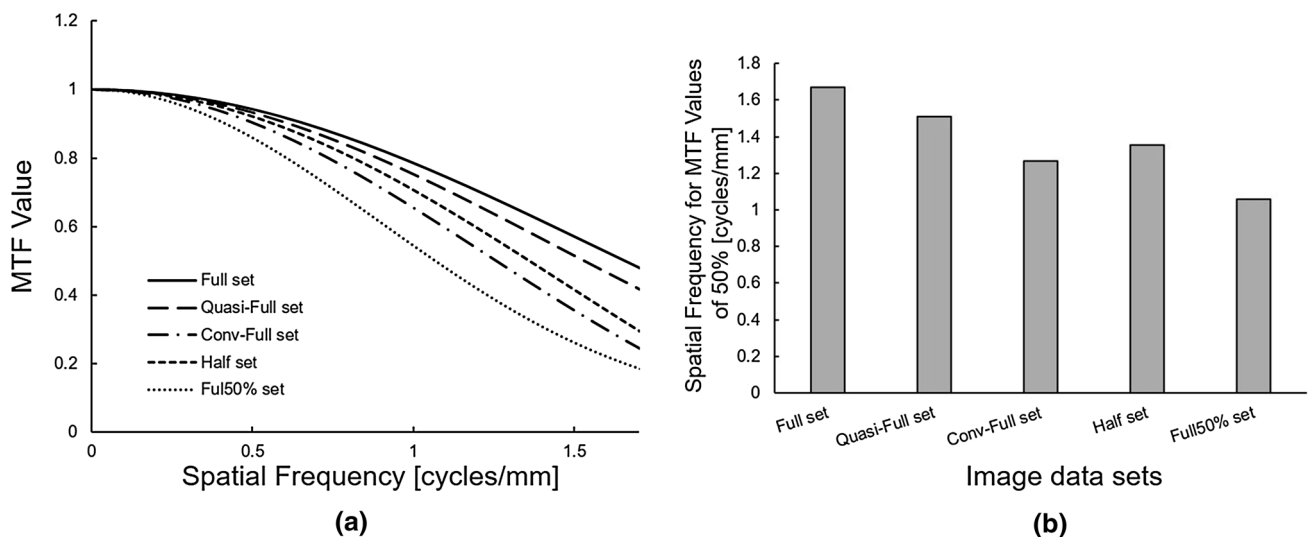


Fig. 6 Results of the modulation transfer function (MTF) in each data set. **a** MTFs obtained using the Full set, Shift-Linear method (Quasi-Full set), Conv-Full set, Half set and Full50% set. **b** Spatial frequency for MTF values of 50% in each data set

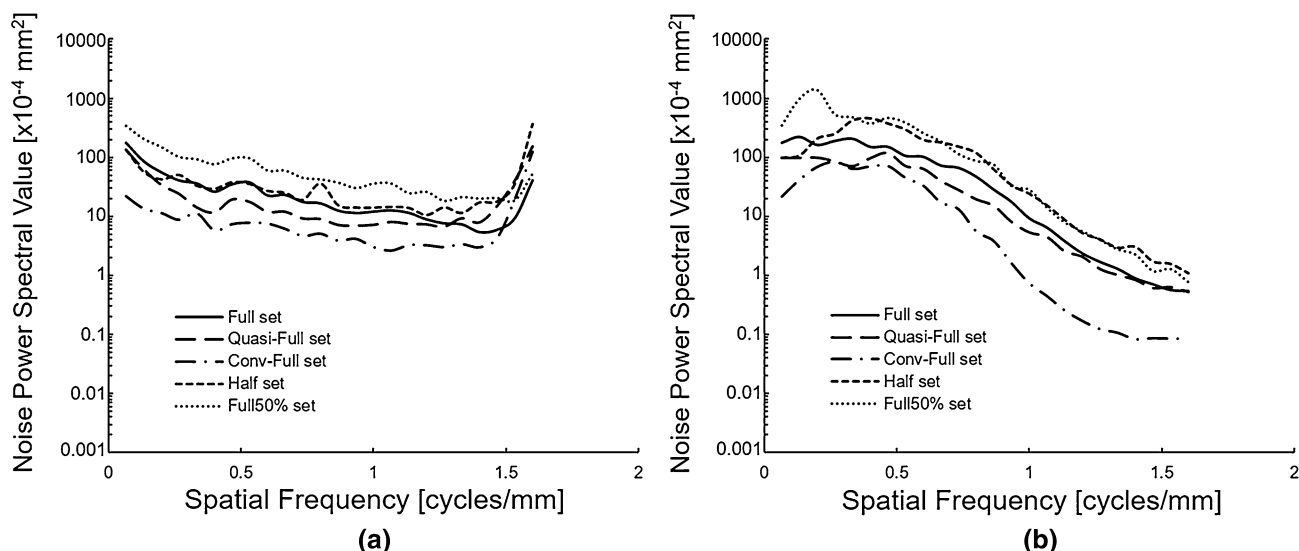
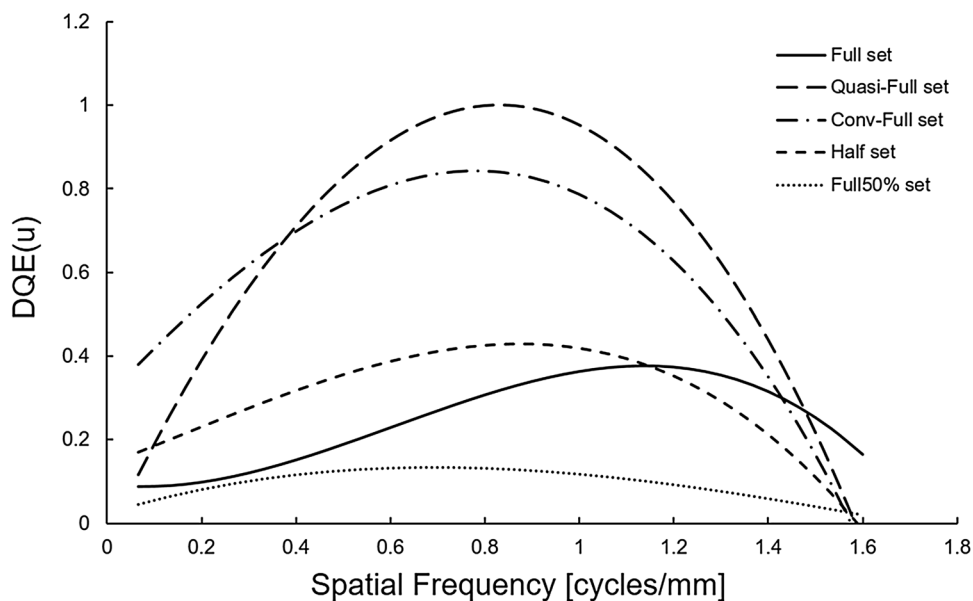


Fig. 7 Results of noise power spectrum (NPS) in each reconstructed condition. **a** u -axis; **b** v -axis. The v -axis is the travel direction of the X-ray tube. The NPS of the Shift-Linear method set (Quasi-Full set) was improved compared with that of the Conv-Full, Half and Full50% sets

Fig. 8 The detective quantum efficiency ($DQE(u)$) of each data set. The $DQE(u)$ values of each data set were normalized using maximum $DQE(u)$ value of Quasi-Full set. The reference number of photons entering the X-ray detector was employed that of the Full set. The number of photons used for the other data sets were estimated using the number of projections or the relative exposure dose. The $DQE(u)$ of the Quasi-Full set was superior to that of the Conv-Full set in range of spatial frequency between 0.4 and 1.5 cycles/mm



found to be superior to those of the Full and the Conv-Full sets by taking into account the estimated numbers of photons entering the X-ray detector.

The CNR values obtained for each data set are shown in Fig. 9. As shown, the CNR obtained for the Conv-Full set was superior to those of the Full and the Quasi-Full sets, and there was no significant difference between the CNRs of the Quasi-Full and the Full sets. Additionally, the CNRs of the Half and Full50% sets were inferior to those of the Full and Quasi-Full sets.

4 Discussion

In this study, the image quality of the tomosynthesis images reconstructed with half the number of projection images and the interpolation images acquired using the Shift-Linear method was only slightly degraded. By taking into account the shifted distance between two pixels for pixel linear interpolation, image blurring resulting from pixel shift could be significantly reduced compared with

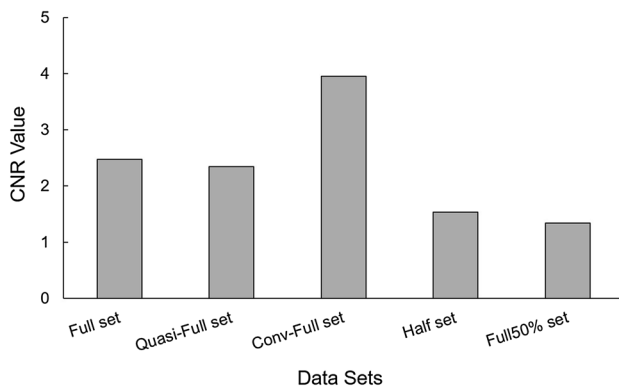


Fig. 9 The contrast-to-noise ratio (CNR) values of the reconstructed tomosynthesis image using each data set at the in-focus plane. The CNR estimated from the Quasi-Full set was approximately equal to that of the Full set. The CNRs estimated from the Half set and Full50% sets were inferior to that of the Quasi-Full set

that of the conventional linear interpolation method. The MTFs of the Half and Full50% sets were inferior to those of the Full and the Quasi-Full sets because of the increased image noise present in those data sets [17].

The NPSs of the Half set were degraded because of the decrease in image data used for reconstruction. Additionally, the NPS obtained with the Full50% set under low-dose conditions was degraded because the exposure dose per projection image was reduced to 50% of that of the Full set.

Although the obtained values of MTF and NPS of the Quasi-Full set were comparable with those of the Full set, the $DQE(u)$ of the Quasi-Full set was clearly superior to that of the Full set. This result highlights the potential utility of our proposed method.

The DQE of the Conv-Full set was also superior to that of the Full set; however, the improvement in the noise property of the Conv-Full set resulted from image blur. Therefore, the conventional interpolation method was not adequate for the interpolation of tomosynthesis projection images. For the same reason, the CNR value of the Conv-Full set was higher than those of other data sets; the changes in CNRs were related to the image noise of the reconstructed tomosynthesis image rather than an image intensity of the signals on the image [18]. However, the CNR value obtained from the Quasi-Full set was comparable with that of the Full set. When we comprehensively assessed image quality, we must conclude that the tomosynthesis images reconstructed using half the number of projection images complemented the interpolation images obtained using the Shift-Linear method were capable of maintaining image quality despite the reduced number of projection images and the consequent reduction in exposure dose.

In this method, depth information was also maintained by estimating each shifted distance of the pixel location between two projection images on, before, and after

the interpolation phase. A degradation of the MTF of the interpolation image could be observed in general as result of inaccurate estimation of the shifted distance. This was caused by the fact that pixels corresponding to the same location in different input images had different pixel values. However, in this study, we attempted to minimize the variation in pixel values of two projection images to as small as possible using pre-image processing to normalize pixel values using the first projection image. Furthermore, using a template-matching technique with only the Y -direction scan corresponding to the travel direction of the X-ray tube, the accuracy of the shifted distance estimation could be considerably improved.

The proposed method, however, has limitations. The kernel size used for template matching was not optimized, and must be modified according to the input images. Reducing the kernel size will reduce the calculation time, but an insufficient kernel size will result in pixel mismatch and imperfect shifted distance estimation. Two examples of the interpolation images, which were created using kernel sizes of 3×3 and 21×21 are shown in Fig. 10a, b, respectively. When using the 3×3 kernel size, there were many pixels missing due to insufficient kernel size for searching the target pixel between two input images. On the other hand, when the 21×21 kernel size was used, the interpolation was successfully achieved. In the present study, we empirically estimated a suitable kernel size a priori so that the resulting interpolation image would not have unmatched pixels. Therefore, to accurately apply this method in clinical situations, it is necessary to determine a suitable kernel size for each individual case.

There was another limitation in the reconstruction technique for tomosynthesis imaging. To reduce the exposure dose and image noise, the iterative reconstruction (IR) method would be the first choice for reconstructing tomosynthesis images. However, the IR algorithms included in the majority of tomosynthesis devices are not published in detail and, therefore, we could not replicate them outside of the tomosynthesis equipment. Because the majority of IR applied a frequency filter function for all reconstructed images, the MTF measured from the IR-reconstructed image had a maximum value of more than 1.0. In addition, the relationship between image noise quality and the degree of dose reduction was not consistent in the IR. Therefore, we did not use the IR method in this study.

For indicating actual changes in the clinical images, a sample set of the clinical images reconstructed using the Full, Quasi-Full, Conv-Full, and Half sets are presented in Fig. 11. The tomosynthesis images reconstructed using the Conv-Full and the Half sets exhibited a significant degradation of image quality, whereas the reconstructed image obtained from the Quasi-Full set with the proposed method was comparable with that of the Full set. This shows that,

Fig. 10 Examples of interpolation images using two types of kernel size 3×3 (a), and 21×21 (b) in the template-matching method at a height of 90 mm above the table. Multiple blank pixels occurred in a due to the missing data in template matching. (Patient data: male, 81 years of age, post-total hip arthroplasty. Parameters: 75 kV, 300 mA, 10 ms, 4.585 mSv [entrance dose])

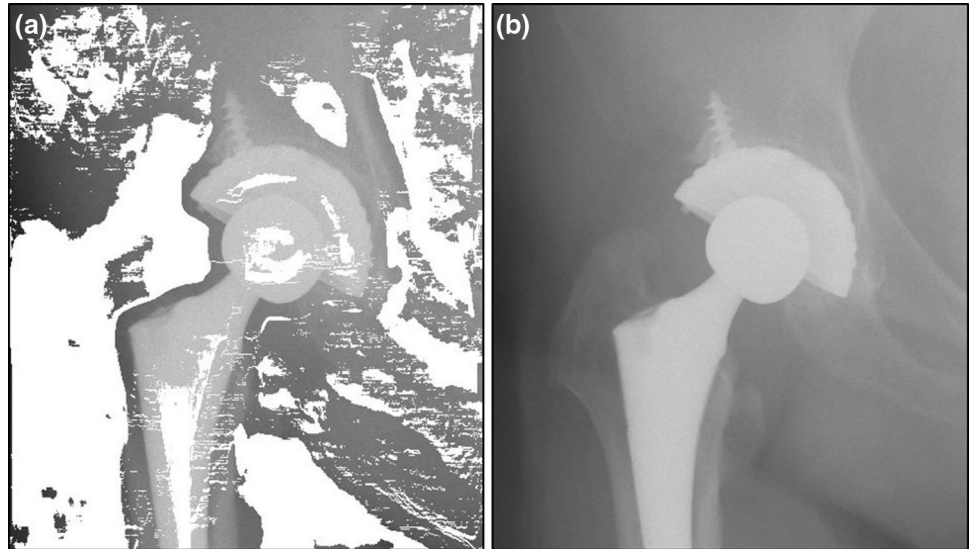
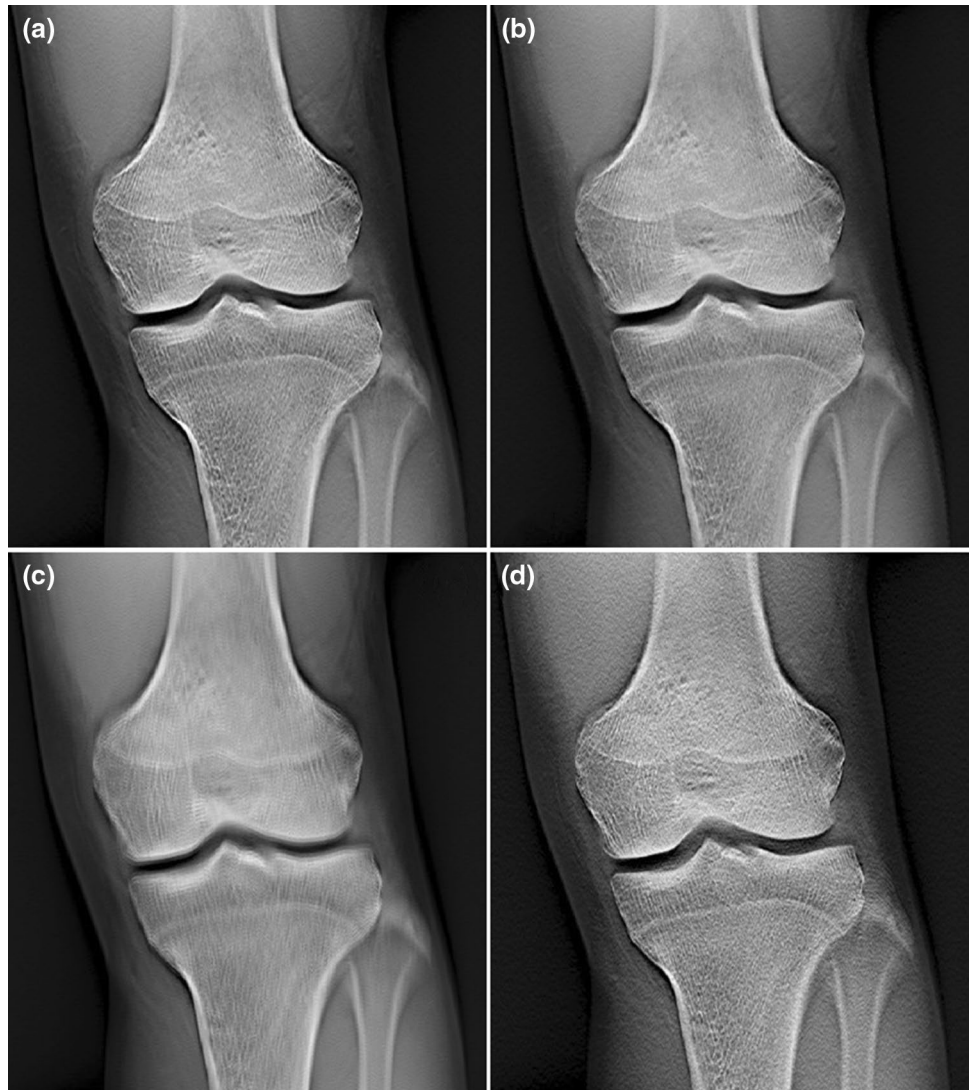


Fig. 11 Clinical tomosynthesis images reconstructed using Full set (a), Half with Shift-Linear method set (b), Conv-Full set (c), and Half set (d) at a height of 80 mm above the table. (Patient data: male, 40 years of age, no abnormality. Parameters: 65 kV, 200 mA, 12 ms, 2.371 mSv [entrance dose])



using the interpolation images obtained using the Shift-Linear method, we could maintain the quality of the clinical image with an approximately 50% reduction in the number of real projections.

5 Conclusion

In this study, we proposed a new method to reduce the number of real projections required to reconstruct a tomosynthesis image by replacing some projection images with interpolation images obtained using a modified conventional linear interpolation technique. The results of the image quality evaluation demonstrate that the proposed method is capable of preserving image quality, despite using a reduced number of real projection images compared with that of the full projection images. In conclusion, this method effectively decreased the number of real projection images necessary.

Compliance with ethical standards

Conflict of interest The authors declare that they have no conflict of interest.

Statement of human rights The Ethics Committee for the Faculty of Medicine, Tottori University, approved the use of the clinical images for this retrospective study (Approval number: 1601A110). Formal consent is not required for a study of this type at the authors' institution.

Statement of animal rights This article does not contain any studies with animals performed.

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