Performance characterization of microtomography with complementary metal-oxide-semiconductor detectors for computer-aided defect inspection

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We developed a computer-aided defect inspection system based on computed tomography (CT). The system consists of a homemade small cone-beam CT (CBCT) system and a graphical toolbox, which is used to extract a computer-aided design (CAD) model from the CT data. In the small CBCT system, the x-ray imaging detector is based on a complementary metal-oxide-semiconductor photodiode array in conjunction with a scintillator. Imaging performance of the detector was evaluated in terms of modulation-transfer function, noise-power spectrum, and detective quantum efficiency. The tomographic imaging performance of the small CBCT system was evaluated in terms of signal-to-noise ratio and contrast-to-noise ratio. The graphical toolbox to support defect inspection incorporates various functional tools such as volume rendering, segmentation, triangular-mesh data generation, and data reduction. All the tools have been integrated in a graphical-user interface form. The developed system can provide rapid visual inspection as well as quantitative evaluation of defects by comparing the extracted CAD file with the original file, if available, of an object. The performance of the developed system is demonstrated with experimental CT volume data. © 2009 American Institute of Physics. [DOI: 10.1063/1.3124360]

I. INTRODUCTION

X-ray computed tomography (CT) is not only a well recognized imaging modality in medicine1 but also an essential one in industrial quality control and nondestructive evaluation.2,3 The recent cone-beam CT (CBCT), enabling three-dimensional (3D) scanning in almost real time, pushes further up the necessity in industry. Although the detailed technologies in CBCT, such as imaging detectors, scanning geometry, and image reconstruction algorithms, are still under development, the overall framework has been matured enough for diverse applications. Recent developments in microfocus x-ray sources and large-area pixel detectors resulted in the development of CBCT systems with resolving power of the order of tens of micrometers; these CBCT systems are called microtomography (micro-CT) systems.4,5 Although micro-CT has the same technology as conventional CT, and thus, not a novel technology, the micro-CT has many new potential applications in biological as well as materials sciences.6–9

Moreover, with the 3D volume CT, the output voxel data can be used for rapid prototyping (RP) modeling, nondestructive testing and evaluation, and reverse engineering, etc. In these aspects, the extraction of computer-aided design (CAD) models of an object from CT data is essential.

IV. MATERIALS AND METHODS

A. Micro-CT system

We developed a cost-effective laboratory micro-CT system as well as software to support the defect inspection of small parts. The core of the micro-CT system is an x-ray detector. Since the performance of the detector is mostly responsible for image quality and eventually the quality of the tomographs, characterization of the detector is an important issue. The imaging characteristics of the detector were evaluated in terms of modulation-transfer function (MTF), noise-power spectrum (NPS), and detective quantum efficiency (DQE). Tomographic imaging performances of the system, such as signal-to-noise ratio (SNR) and contrast-to-noise ratio (CNR), have also been evaluated by using quantitative phantoms. The micro-CT system and software, which is the integrated graphical toolbox for extracting CAD files in standard triangulation language (STL) format from the voxel data, are described in detail. The overall performance of the developed toolbox is demonstrated with the experimental volume CT data.
The CMOS photodiode array has a format of 1024 pixels with a pitch of 48 μm. Only one narrow side of the CMOS photodiode array incorporates the readout electronics to allow tiling of the CMOS photodiode array into large mosaics. In this study, two CMOS photodiode arrays were tiled side by side and therefore the FOV was about 50 × 50 mm². Two parallel voltage signals from the two CMOS photodiode arrays are then simultaneously digitized to 12-bit gray level through dual analog-to-digital converters (ADCs) (PCI-6111, National Instruments, USA). The 12-bit image data are converted into 16-bit form for postprocessing and image reconstruction.

The frame time of the CMOS detector to acquire a single projection could be variable and we normally operated the detector with the frame time of 275 ms. Algorithms and software to acquire digital x-ray images, to apply postimage processing, and to reconstruct volumetric images were developed by using Microsoft Visual C++. When we operated the system in the CT scan mode, the magnification factor was usually set to about 2 and image data were obtained through a 360° scan. For image reconstruction, we employed the Feldkamp’s cone-beam algorithm to the projection data filtered with a Ram-Lak filter (or high-pass filter). The Feldkamp’s algorithm is a simple extended version of the conventional filtered backprojection method in longitudinal direction by considering cone angles. Therefore, it is an approximate approach and may cause distortion when the cone angle is wide. However, since Feldkamp’s algorithm is very fast and easily applicable, it is mostly widely used in commercial CBCT systems. For a typical scan in this study, the size of the acquired data was about 1 Gbyte (=2 bytes/pixel × 1024² pixels/projection × 500 projections).

**B. Imaging characteristics of the CMOS detector**

The performance of the CMOS detector has been evaluated in terms of MTF, NPS, and DQE, which are fundamental metrics describing the physical imaging characteristics of an imaging system. The MTF describes the contrast transfer efficiency as a function of an object detail, hence the resolving power of an imaging system. The NPS measures the change in the noise amplitude as a function of spatial frequency and bridges the noise power and spatial resolution in an image. The DQE is a measure of the square of the SNR (SNR²) transfer and measures the fraction of the incident x-ray fluence contributing to image quality. Therefore, the DQE represents the performance of an imaging system. A detailed description of measurement procedures can be found in a previous study. Briefly, the presampling MTF was measured using the slanted-slit method to avoid aliasing and the NPS was determined by 2D Fourier analysis of white images. Gain and offset corrections were performed on all images used to obtain the MTF and NPS. We actually measured a normalized NPS (NNPS) to avoid the additional measurement of the detector gain. A 45 kVp x-ray spectrum tailored by a 0.5-mm-thick aluminum filter was used for all measurements. The DQE was then assessed from the measured MTF, NPS, and the estimated photon fluence \( \bar{q} \) as

\[
DQE(f) = \frac{MTF²(f)}{\bar{q} \times NNPS(f)},
\]

where \( f \) denotes the spatial frequency. The fluence was calculated using the experimentally measured exposure and half-value layer, and the computational program for x-ray spectral analysis. The exposure rate at the entrance surface of the detector was measured by replacing the detector with a calibrated ion chamber (Victoreen 6000–528, Inovision, USA) while keeping the same distance.
C. Tomographic imaging performances of the micro-CT system

Tomographic imaging performance of the developed micro-CT system was evaluated with homemade quantitative phantoms as shown in Fig. 2. A water-filled cylindrical acrylic vessel having a diameter of 30 mm was used in evaluating the voxel signal and noise characteristics. The CNR was evaluated with the contrast phantom, which consists of six low-contrast inserts. The contrast phantom is completed when the inserts are immersed in the water vessel. As shown in Fig. 2(b), each insert was made of commercial electronic density phantoms (Model 76-430, Nuclear Associates, NY, USA) and it had three different diameters; 1.5, 3.0, and 5 mm. The insert materials are plastic water (1.03 g/cm³), nylon (1.15 g/cm³), polyethylene (0.95 g/cm³), acryl (1.18 g/cm³), polystyrene (1.11 g/cm³), and polycarbonate (1.18 g/cm³), and their densities are similar to that of water. From the cross-sectional images obtained with the contrast phantom, the CNR was calculated by

\[
\text{CNR}_{a} = \frac{\bar{S}_a - \bar{S}_w}{\sqrt{\sigma_a^2 + \sigma_w^2}},
\]

where \(\bar{S}\) and \(\sigma\) are the mean signal and standard deviation of the voxel values, respectively. The subscripts \(i\) and \(w\) denote the inserts and the background water region in the contrast phantom, respectively. The CNR was analyzed in terms of various imaging conditions, such as dose, diameter of inserts, and phantom, respectively. The CNR was evaluated with the contrast phantom, which contains water, is scanned for the estimation of the voxel noise and SNR. Each insert has three different diameters: 1.5, 3.0, and 5.0 mm.

D. Graphical toolbox to extract CAD files from CT data

In this study, we used two approaches for defect inspection. One is a simple visual inspection of three-dimensionally reconstructed image data. The other is the defect inspection based on CAD data. In these regards, we developed a graphical toolbox capable of visualizing 3D data as well as extracting CAD files from the CT data. Main functional tools incorporated in the developed toolbox are the volumetric rendering of the reconstructed 3D data, the segmentation of region-of-interests (ROIs), the generation of triangular-mesh data of the segmented ROIs, and the smoothing and reduction of mesh data.

Volume rendering is a discrete representation and visualization of objects as sampled data, of not only the surface but also the entire inner of an object, in three dimensions. We independently considered two different methods, such as ray casting and maximum intensity projection (MIP). Ray casting is a direct volume rendering that sums the intensity values of voxels based on the light absorbed as it propagates from back to front through the 3D voxel data set. On the other hand, in the MIP method, the shade of gray of a voxel having maximum intensity is chosen in the 3D voxel data encountered along the projection line.

We consider the STL format as a CAD file, which is mesh data containing geometrical information that details the object. It consists of vectors designating three vertices of a triangle and its surface normal vector. The procedure to extract a STL file from the voxel data mainly consists of two steps: segmentation of voxel data in binary form and triangulation of the segmented volume data. Segmentation, the most important procedure, is very sensitive to each voxel value and largely responsible for the reliability of the mesh data. We basically incorporate the thresholding method based on histogram analysis. The threshold is determined by a simple calculation that maximizes the variance of the intensities between the user-defined ROI and the other regime in the histogram. This approach can be easily realized with effective computation.

In order to generate triangle meshes, we employed the well-known marching cube algorithm. A virtual cube is defined by eight-pixel (or eight-vertex) values between two vertically consecutive slices of the segmented volume data. Therefore, this algorithm determines how the slice surfaces intersect the cube by comparing the vertex values, which are prescribed as units, with the segmented data. Since possible triangles (256 cases) are already prepared as a look-up table, the triangulation is automatic. The cube marches to the next location and the process is repeated. However, since the number of triangles that could be generated per cube is limited to four, the degree of surfaces that should be expressed is restricted. Reducing the gap between slices or a higher-order interpolation method might be a way to overcome this restriction, but it entails computational cost. It is completed when the triangulated meshes are exported according to the STL file format.

Since the file size of the mesh data extracted by the marching cube algorithm is normally large, it is often necessary to remove unnecessary meshes and merge the meshes in...
coplanar planes. This data reduction process should preserve the original topology and approximate the original geometry reasonably. We employed various mesh-data reduction algorithms, such as mesh decimation, quadric decimation, and quadric clustering.

Most of the algorithms describing each tool were coded by using Microsoft Visual C++. For user convenience, the toolbox was developed in an integrated graphical user interface (GUI) form.

E. Experimental simulation

In order to implement the developed toolbox, we designed a chessman, bishop, in STL file format using a commercial CAD software, Pro/Engineering™ and manufactured an RP sample using optical stereolithography (SLA 350, 3D Systems, USA) with SL-51902 resin. The manufactured RP bishop sample was then scanned by the developed micro-CT system and a 3D image was reconstructed in voxel size of 0.096 × 0.096 × 0.096 mm³. By using the developed toolbox, we extracted the STL file from the scanned CT voxel data and used the developed toolbox to compare the extracted STL file with the original one. The bishop sample was designed to have a height of 67.44 mm and the manufactured RP sample was measured to 67.58 mm. Both the imaging FOV of the CMOS detector and the introduction of magnification in imaging geometry restricted the sample height to be imaged to about 34.69 mm.

III. RESULTS

A. Radiographic imaging performance

Fourier analysis of the radiographic imaging performance of the detector equipped in the micro-CT system is summarized in Fig. 3. The measured MTF is plotted in Fig. 3(a). At 10% of MTF, the spatial resolution is about 10.5 mm⁻¹. We could easily resolve 10 lp/mm of the high-resolution line-pair phantom from a separate measurement as shown in Fig. 4.

Figure 3(b) describes the NNPS as a function of the spatial frequency. There is a noticeable increase in noise power around zero frequency due to large-scale gain variation across the detector, which would normally be eliminated by the gain correction procedure. In order to remove or reduce the large-scale nonuniformity, a 2D second-order polynomial fit to an image is used and is subtracted from the corresponding image. In this study, however, we subtracted the mean value of the image instead. The measured NPS shows essentially white-noise property with respect to the spatial frequency. The higher MTF performance might reflect this white-noise property in part because the noise is also transferred through the MTF as a signal. A previous study pointed out that in a detector employing a thin phosphor, the extra noise due to the direct absorption of x-ray photons in the photodiode is white. Together with the measured MTF and NNPS, the DQE was calculated and the result is shown in Fig. 3(c).

B. Tomographic imaging performance

Cross-sectional images of the contrast phantom with respect to various parameters, such as the diameter of the low-contrast inserts, the dose and the slice thickness, are shown in Fig. 5. We applied narrow threshold windows to clearly differentiate the inserts. The default voxel size is 0.139 × 0.139 × 0.278 mm³. The voxel heights or the slice thicknesses of the images shown in Figs. 5(f) and 5(g) are 0.556 and 1.112 mm, respectively.

![Radiograph of the line-pair test phantom, which supports the result of measured MTF.](image-url)
For three different diameters of inserts [see Figs. 5(a)–5(c)], no severe change in contrast was observed, even down to the diameter of 1.5 mm, for each insert material. The acquired images also showed similar CNR values and tendencies as a function of dose for the three different diameters of inserts. For the inserts having a diameter of 1.5 mm, the calculated CNR as a function of the dose at the AOR is plotted in Fig. 6. The CNR was found to be proportional to the square root of the dose.27

Figure 5(a), 5(d), and 5(e) demonstrate the effect of dose on visual impact. In general, the image quality of a radiograph is mainly dependent on the number of detected photons and it is definitely increased as the number of incident x-rays is increased. The tomograph becomes poorer as the dose decreases. For a given dose, the increase in the slice thickness has the same effect as an increased number of detected photons, as shown in Figs. 5(a), 5(f), and 5(g).

In all the measurements, the worst CNR was observed with the insert made of acrylic. The minimum resolvable CNR with the naked eye was approximately 1 and the voxel noise of the water phantom corresponding to this CNR value was about 250 CT numbers.

The SNR, calculated from the measured cross-sectional image of the water phantom, as a function of dose is shown in Fig. 7. Like the CNR, the SNR is also dependent on the square root of the dose. At the dose of 83 mGy, the SNR is about 6, and the corresponding voxel noise is 295 CT numbers.

### C. Graphical toolbox and the extraction of CAD files from CT data

Figure 8 describes the main frame of the developed graphical toolbox. The toolbar at the top of the main frame contains various symbolic buttons, which are used to run the corresponding functions, such as loading of the volume data or STL data, handling of visualization and cropping, the ex-
The approximate Feldkamp’s algorithm for cone angle could produce an error,11 the generated error for small cone angle or small sample volume would be negligible. For the reliable performance evaluation of the developed system, the error should be quantitatively analyzed in terms of each contributing factor.

IV. DISCUSSIONS AND CONCLUSION

The overall MTF of the detector system is affected by various physical parameters such as the x-ray focal spot size, the slit width, the magnification ratio, and the inherent detector resolving power.19 The inherent resolving power of the detector employed in this study is mainly determined by the optical photon scattering in the phosphor screen and the pixel aperture size of the CMOS photodiode array. Since the SDD was set to 200 mm when measuring MTF with a slit camera having a 10-μm-width slit aperture and made of the 1.5-mm-thick tantalum28 and the slit camera was in contact with the entrance surface of the detector, we neglected the magnification of focal spot and slit width. The theoretical MTF due to an aperture function can be expressed by the “sine cardinal” or simply “sine” function. Accounting for the one-dimensional (1D) geometries of the focal spot, the slit aperture, and the detector pixel, the theoretical MTFs are plotted in Fig. 3(a). The pixel MTF was estimated with consideration of the geometric pixel fill factor of 0.87. From the results, the overall MTF was seemed to be mostly determined by the resolving power of the phosphor screen; in other words, the dominant blurring factor was the optical photon scattering within the phosphor screen.

On the other hand, tomography is obtained with a certain magnification factor, i.e., the ratio of SDD to SOD. Therefore, there is a compromise between the magnification of details in an object and the blur due to the magnified focal spot size. In addition, image reconstruction is another blurring factor in tomography.29 In the previous study on the development of a microtomography system incorporating a focal spot size of 10 μm and a pixel pitch of 50 μm,19 the system total MTF was mostly determined by the detector resolving power at the magnification factor of 2. We assumed that the system MTF of our micro-CT system would be similar. However, the MTF in tomography should be measured for optimal and reliable use.

As long as the additive electronic noise is not exceptionally large, the NPS of an imaging system is mainly governed by the stochastic variation of information-carrying quanta introduced in various image-forming states. Typical examples are the interaction of x-rays in a scintillator, the absorption of its energy, the conversion of x-ray to optical photons, the spread of an optical photon burst, and the escape of optical photons, and, in a photodiode, the collection of optical photons, the conversion of optical photons to electronic charges, and the spread of electronic charges. This stochastic variation increases the noise at the output, and whose effect can be seen by examining the NPS. According to Rossmann,26 because the spatial-frequency dependence of the signal transfer is described by $MTF(f)$, the NPS would behave as $MTF^2(f)$. However, as shown in Fig. 3(b), the

FIG. 9. (Color online) Experimental simulation with a chessman, bishop, using the developed toolkit. (a) Original CAD data, which were as an input file for the RP machine, and the manufactured RP bishop sample. (b) 3D-rendered images of voxel data obtained from the CT scan. Two different visualizations, ray casting, and MIP, are shown. (c) Extracted CAD files from the voxel data by using the developed toolkit. The left is the extracted, as is, and the right is after decimation and smoothing processes. (d) A map describing the difference between the original and the extracted CAD files.
NPS of our detector system shows behavior close to that of a white noise spectrum except at the very low frequency band, probably due to the imperfect removal of the fixed-pattern noise during the gain-offset correction procedure. For comparison, a best fit function, proportional to \( \text{MTF}^2(f) \), is plotted in Fig. 3(b). A large discrepancy is observed in the high spatial-frequency region. From Lubberts,\(^{30}\) the NPS generally falls off more slowly at high frequency than the \( \text{MTF}^2 \) because of the stochastic variation caused by the depth-dependent interaction of x rays within a phosphor screen and the fluctuation of processes at the corresponding depth. Badano et al.\(^{31}\) demonstrated the Lubberts’ effect on a phosphor screen by estimating the Lubberts fraction, \( L(f) = \text{MTF}^2(f)/\text{NNPS}(f) \) based on Monte Carlo simulation. However, their result on the phosphor screen having a similar thickness as that employed in our detector is quite different. Our measurement and analysis showed a much stronger additional white noise spectrum except at the very low frequency band, the direct x rays are those that are unattenuated from the phosphor screen and that directly interact with the photodiode. This additional white noise can further raise the spectral density in the high spatial frequency band. The proof that the NNPS due to the direct x rays is white is shown in Ref. 32. Considering this additional white noise, we analyzed the measured NPS with a least-squares fit of the form \( a \text{MTF}^2(f) + b \), where \( a \) and \( b \) are the constants obtained from the numerical fit and plotted the resultant fit in Fig. 3(b). This analysis was directly applied to the DQE result as shown in Fig. 3(c). The analysis reasonably agreed with the measured DQE data. The degradation of NPS due to the direct x-ray absorption noise decreased the DQE performance, especially in the high frequency region.

Not only do direct x rays affect the image quality or the detector performance, such as NPS and DQE, but they also can give rise to a change in the characteristics of the active devices in the photodiode array, such as the leakage current of a photodiode and the threshold voltage of a switching transistor. The radiation-induced increase in the dark signal and noise can result in the gradual reduction in the dynamic range and enhance the fixed-pattern noise in digital radiography.\(^{33}\) Therefore, a phosphor screen must be designed or selected optimally. In digital radiography, the offset and gain correction is typically performed. With carefully updated and applied correction during the operation of CBCT, the detrimental effects of increased dark current on NPS and DQE can be overcome.\(^{33}\)

The applicability of the developed system is somewhat limited by the FOV of the detector. If magnified images and tomograms are needed for a detailed analysis of an object to be scanned, FOV becomes a more serious problem. The detector can extend the tiling approach of the CMOS photodiode array used in this study to a larger dimension, for example, the FOV of about 100x100 mm\(^2\) with 2x4 arrangements of the photodiode array. Moreover, commercial large-area flat-panel detectors are available. With an undersized detector, the extension of the reconstructed volume with mechanical motion or a smart image reconstruction algorithm is another approach that can be used to overcome the restricted FOV of a detector. Spiral motion, which is typical in medicine, on a long object is an example. For a wide object, by shifting a detector laterally so that data from at least half the width of the object are acquired for each view, the image reconstruction based on the estimation of the missing rays through the unprojected part of the object from the redundant rays can be used to enlarge the reconstructed volume in laterally.\(^{34}\)

A series of algorithms to realize computer-aided defect inspection has been integrated in a GUI form. The developed graphical toolkit was used to extract CAD files from the scanned CT volume data of an RP sample. In the serial procedures, the most important procedure is the segmentation. An object containing various inner structures with similar material properties, such as density and atomic number, requires the implementation of more elaborate segmentation algorithms. However, the segmentation algorithm used in this study is reasonable for typical industrial applications because most industrial parts are composed of a few materials.

Although the developed system is somewhat restricted in FOV, it can be used for diverse industrial applications, such as the inspection of solder joints in electronic components and reverse engineering of small parts, and for biological applications, such as small-animal imaging, tissue engineering, etc.

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