

PAPER • OPEN ACCESS

Varying-energy CT imaging method based on EM-TV

To cite this article: Ping Chen and Yan Han 2016 *Meas. Sci. Technol.* **27** 114004

View the [article online](#) for updates and enhancements.

You may also like

- [A data-driven maximum likelihood classification for nanoparticle agent identification in photon-counting CT](#)
Sumin Baek and Okkyun Lee
- [Sparsity-regularized image reconstruction of decomposed K-edge data in spectral CT](#)
Qiaofeng Xu, Alex Sawatzky, Mark A Anastasio et al.
- [Statistical image reconstruction from limited projection data with intensity priors](#)
Essam A Rashed and Hiroyuki Kudo

Varying-energy CT imaging method based on EM-TV

Ping Chen and Yan Han

National Key Laboratory for Electronic Measurement Technology; Key Laboratory of Instrumentation Science & Dynamic Measurement, North University of China, Taiyuan, People's Republic of China

E-mail: pc0912@163.com

Received 9 November 2015, revised 8 January 2016

Accepted for publication 31 March 2016

Published 20 September 2016



Abstract

For complicated structural components with wide x-ray attenuation ranges, conventional fixed-energy computed tomography (CT) imaging cannot obtain all the structural information. This limitation results in a shortage of CT information because the effective thickness of the components along the direction of x-ray penetration exceeds the limit of the dynamic range of the x-ray imaging system. To address this problem, a varying-energy x-ray CT imaging method is proposed. In this new method, the tube voltage is adjusted several times with the fixed lesser interval. Next, the fusion of grey consistency and logarithm demodulation are applied to obtain full and lower noise projection with a high dynamic range (HDR). In addition, for the noise suppression problem of the analytical method, EM-TV (expectation maximization-total Jvariation) iteration reconstruction is used. In the process of iteration, the reconstruction result obtained at one x-ray energy is used as the initial condition of the next iteration. An accompanying experiment demonstrates that this EM-TV reconstruction can also extend the dynamic range of x-ray imaging systems and provide a higher reconstruction quality relative to the fusion reconstruction method.

Keywords: CT imaging, variable energy, high dynamic, iteration reconstruction, complicated structural component

(Some figures may appear in colour only in the online journal)

1. Introduction

Industrial computed tomography (CT) can be used in the non-destructive testing of safety-critical workpieces [1–3]. However, in an x-ray CT imaging system, because of the limitation of the ray crystal conversion efficiency, charge capacity, photoelectric conversion efficiency and A/D, the dynamic range of the imaging system is limited. Thus, for complicated structural components, the full projection cannot be captured with the conventional fixed-energy imaging method, because the wider thickness range of the components along the orientation of the x-ray penetration exceeds the limit of the dynamic range of the imaging system. In these incomplete projection images, there are regions of overexposure (thinner

areas with high ray energy) and underexposure (thicker areas with low energy) [4, 5]. Conventional CT reconstruction algorithm can't get the construct of these complicated structural components in the limitation of the dynamic range [6].

Processing algorithms exist for 3D CT imaging of complicated workpieces, such as projection offsetting estimation, limited-angle CT, CT image correction, and other processes. However, the rationality of information estimation and the effectiveness of post-processing will affect the correctness of CT information because of the projection information missing at every imaging energy. In addition, with the more complicated construct, the projection of fixed-energy is seriously lacking; as a result, the data offsetting and post-processing cannot ensure effective CT reconstruction [7, 8]. Thus, for the current imaging system with a low dynamic range, the development of a highly dynamic DR/CT imaging method is meaningful for characterizing complicated workpieces.



Original content from this work may be used under the terms of the [Creative Commons Attribution 3.0 licence](https://creativecommons.org/licenses/by/3.0/). Any further distribution of this work must maintain attribution to the author(s) and the title of the work, journal citation and DOI.

For fixed-energy imaging, invalid areas exist, but there are still valid irradiations, for which the thickness is matched to the fixed x-ray energy. By using image sequence fusion, we can obtain the all projection information via multi-energy imaging. Multi-energy imaging, such as dual-energy CT, multi-spectrum based on a photon counting detector, can increase the energy spectrum dimension for CT reconstruction, thereby improving the CT quality [9, 10]. However, these methods mainly consider material distinction and capture different attenuation relative to the x-ray energy by energy subtraction. For complicated structures, projection is incomplete when using the fixed x-ray energy; thus, we must expand the dynamic range of the imaging system. These multi-energy imaging methods cannot be used in reconstruction directly. Here, based on the local efficiency at a fixed energy for complicated structures, we capture the projection sequences with varying x-ray energy at one rotation angle. The other angles are similarly processed. Subsequently, research fusion and a reconstruction algorithm of high dynamic range are used to obtain the CT image [11]. Also in order to solve noise suppression problem in fusion because of weighting coefficients, EM-TV iteration reconstruction is used.

2. Varying energy CT imaging

2.1. Analysis of conventional fixed-energy CT

For the workpiece depicted in figure 1, the thickness of the specimen exhibits a notable difference. The bottom is solid, the centre is hollowed out, and the top is irregular. Image sequences were captured from 60 kVp to 100 kVp with the step of 10 kVp at every projection angle using a 12-bit imaging system, which consists of x-ray source (GE ISOVOLT 450 KV) and detector (VARIAN PaxScan2520) (figure 2). Also in the experiment, the current is set as 1.5 mA, the pixel pitch is 0.127 mm, the integration time is 0.9 s, and the number of frame averaged 8 frames. This CT imaging system has offset correction and gain correction. So the acquires images are presented in figure 3.

As shown in figure 3, because the variations in the effective thickness of the component along the direction of x-ray penetration exceed the limit of the dynamic range of the x-ray imaging system, the image sequences exhibit only a partial response. Also in the higher energy, because of the initial gain correction coefficient has changed, there are some strip-like noise in the projection. So the CT results based on direct reconstruction are shown in figure 4.

As shown in figure 4, we cannot obtain the full construction information at any one energy, i.e. low energy cannot reveal the inner information, and high energy cannot show the outer edge.

2.2. Varying-energy CT imaging

In x-ray imaging, if all other scanning parameters of the x-ray imaging system are fixed, varying the current will not change



Figure 1. Images of a complicated test specimen.

the distribution of the x-ray energy spectrum but will increase the initial intensity. Thus, varying the voltage changes the distribution of the spectrum. When the tube voltage is higher, the penetration depth will be greater and the density suitable for imaging will be higher. As a result, for complicated structural components, the image information will be lacking with a fixed tube voltage because the restricted dynamic range of the imaging system creates a mismatch between the x-ray energy and the effective thicknesses of certain regions of the component. However, in any energy image, we can also obtain the effective local information. Therefore, variable-energy imaging can be an effective approach to the imaging of components with complicated structures. In this manner, all of the projection information can be captured through variable-energy imaging at every projection angle of the CT rotation. The specific principle is illustrated in figure 5.

In figure 5, tube voltages are adjusted from low to high voltages and image sequences were obtained at every projection angle. There are three cases, detailed as follows.

- (1) High voltage: Only effective local information corresponding to the larger effective thicknesses will be obtained, but at the areas of smaller effective thickness, because the number of radiation photons will exceed the acceptance tolerances of the detector, the photons will overflow to pollute the circumjacent pixels. This pollution produces overexposure.
- (2) Low voltage: Only the effective local information corresponding to the smaller effective thicknesses will be obtained, but at the areas of larger effective thickness, because the penetrating strength of the ray photons is smaller, there are fewer accepted photons and underexposure will occur.
- (3) Medium voltage: Only the effective local information corresponding to the areas where the tube voltage and the effective thickness are well matched will be obtained. However, for the larger or smaller effective thicknesses, both underexposure and overexposure will occur.

Based on the above-described situations, with increasing voltage, areas with greater thickness or higher density of the complex components are gradually shown in the corresponding x-ray image. Next, we apply image processing to fuse this sequence and reconstruct the image of the complex components. However, taking the unknown internal structure into account, to ensure the continuity of the adjacent energy images, the tube voltage must be gradually increased with the smaller voltage interval.

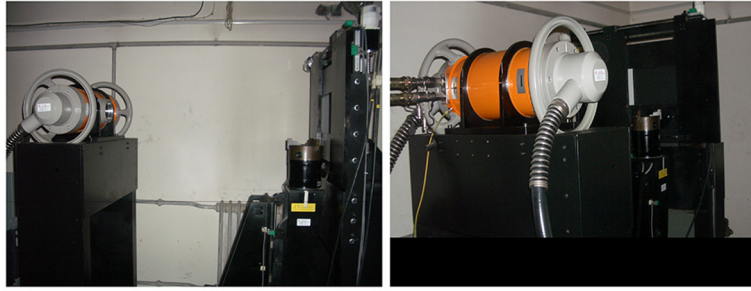


Figure 2. The experiment CT system.

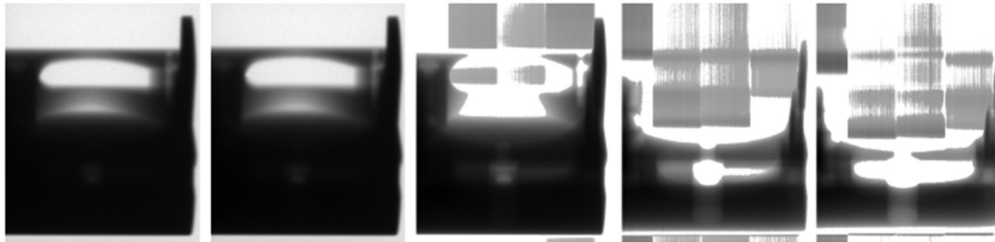


Figure 3. The image sequences collected at varying energies over an energy range from 60 kVp to 100 kVp at the projection angle of 90°.

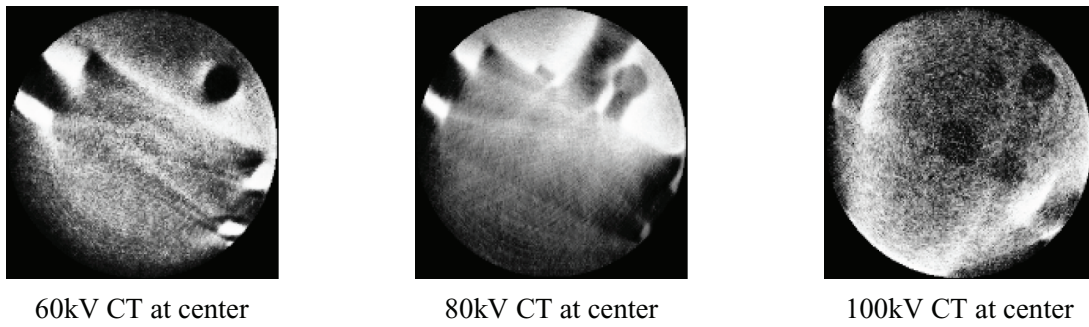


Figure 4. Fixed-energy CT images at different energies and different part.

3. High dynamic fusion

3.1. Effective fusion area

Actually, because of imaging characteristic and SNR (signal to noise ratio) of the detector, the effective gray range is less than the dynamic range of detector. Assuming the dynamic range of the flat detector is $[0, D]$, if the x-ray energy matches the detection object, $[0, D]$ will represent well the gray information of the object. However for the object with wide thickness ranges, there will also be an underexposed area $[0, D_{\min}]$ and an overexposed area $[D_{\min}, D_{\max}]$ at the single energy, where the image quality is poorer. Therefore, the effective information that matches the single energy is only $[D_{\min}, D_{\max}]$. Here, $[D_{\min}, D_{\max}]$ is called the best gray scope. For our imaging system 12-bit, the experiment analyzed the image average gray values about every steel plate, when achieving optimal imaging. The image quality is guaranteed by IQI (image quality indicator). The best gray scope is defined as $[1000\ 1800]$. So we can get the processed image sequences.

$$X_i = \begin{cases} I_i & I_i \in [D_{\min}, D_{\max}] \\ 0 & I_i \notin [D_{\min}, D_{\max}] \end{cases} \quad (1)$$

In equation (1), the nonzero area of X_i is the effective fusion area, I_i is the directly captured image sequence with different ray energy. Also in order to exclude the strip-like noise, the recursion extraction method is used. The method is based on the gradual changing of image structure. In the lower energy image, these noise areas are the effective areas. Thus, when extracting information, the extraction area of the lower energy image is labeled, and this area will be subtracted from the higher energy image [5].

3.2. Fusion coefficient

As shown in figure 2, the object's information gradually appears in the energy image sequence from thinner to thicker areas. We can fuse these images by linear weighting.

$$X = \omega_1 X_1 + \omega_2 X_2 + \dots + \omega_n X_n \quad (2)$$

where X_i ($i = 1, 2, \dots, n$) is energy sequence from low energy to high energy, X is the fusion image, and ω_i is the weighting coefficient. If no constraint of dynamic range exists, to ensure normal imaging of the entire construct, the imaging energy should be same with X_n . Thus, the thickness is lower and the image is greyer. Therefore, in fusion, $\omega_n = 1$, sequence

X_1, X_2, \dots, X_{n-1} is, respectively added to X_n with the corresponding coefficient ω_i . The relationship of ω_i should be diminishing because the grey of the thinner thickness is greater in the imaging system with no dynamic range constraint.

$$\omega_1 > \omega_2 > \dots > \omega_n = 1 \quad (3)$$

As shown in equation (3), the noise in sequence will be amplified by a coefficient larger than 1, which is higher than in fixed-energy imaging, and impact CT reconstruction quality. To solve this problem, reducing grey weighting cardinality and the coefficient of low-energy imaging is the only effective method of decreasing highly dynamic projection noise. Based on Beer's law, logarithm demodulation can narrow the dynamic range and reduce noise by diminution weighting cardinality. In addition, with compression of dynamic range by logarithm transformation, the projection detail will be more obvious. Thus, we can define the new fusion equation as follows:

$$p = v_1 p_1 + v_2 p_2 + \dots + v_n p_n \quad (4)$$

where p_i ($i = 1, 2, \dots, n$) is the logarithm of X_i , p is the fusion image of the logarithm demodulation sequence, and v_i is the new weighting coefficient. From equation (3), $v_n = 1$, sequence p_1, p_2, \dots, p_{n-1} is, respectively added to p_n with the corresponding coefficient v_i . However, based on the relationship between p_i and X_i ,

$$p_i = \ln \frac{X_{0i}}{X_i} \quad (5)$$

If the dynamic range is unconstrained, then the background grey X_{0i} is certain. If the thickness of object is thinner, then the image grey X_i is higher and the logarithm p_i is lower. Thus, for the variable-energy sequence X_i , the corresponding thickness is gradually increasing. In this way, the logarithm of the low-energy image should multiply the minor coefficient, namely,

$$v_1 < v_2 < \dots < v_n = 1 \quad (6)$$

These coefficients less than 1 can further decrease noise. In dynamic range fusion, we should calculate v_i . Logarithm demodulation changes the grey value but does not change image structure.

3.3. High dynamic range (HDR) fusion based on gray consistency

In addition, in sequence capturing, because a smaller voltage interval is used, the same area will exist in two adjacent frames. In fact, the grey in the same area should be basically the same. As a result, we can use goal optimization to solve for the weighting coefficient. Assume the area is Ω_i between p_i and p_{i+1} , and define the distance between them as follows:

$$d(p_i, p_{i+1}) = \sqrt{\sum_{j \in \Omega_i} (v_i x_{ij} - v_{i+1} x_{i+1,j})^2} \quad (7)$$

Equation (7) can show the grey consistency of p_i and p_{i+1} in area Ω_i , x_{ij} is the gray of j -pixel in X_i . The lower the value of $d(p_i, p_{i+1})$, the higher is the consistency. Thus, we can select v_i to minimize the distance, which can ensure a grey consistency

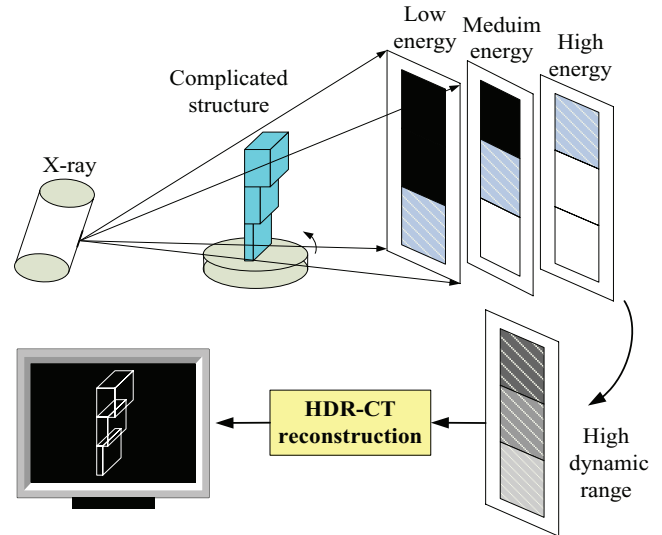


Figure 5. The principle of HDR-CT imaging. At every rotation angle, get multi-energy sequences, which are responding to different thickness. Then by CT reconstruction to get the full construct information.

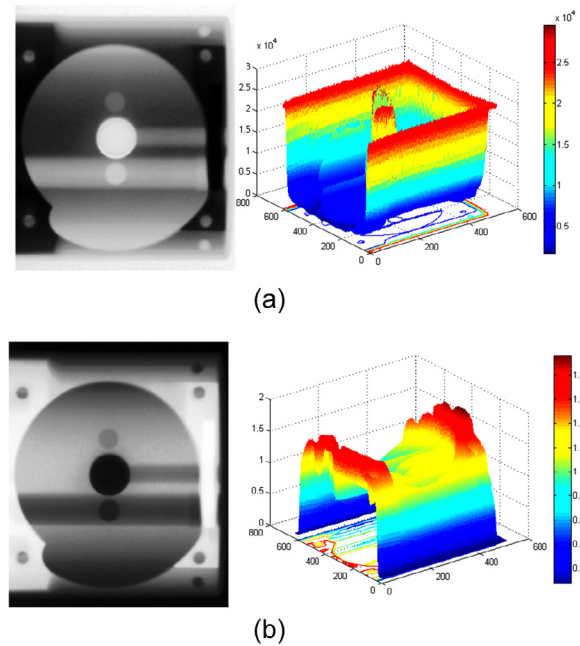


Figure 6. Projected data and 3D grey curve: (a) fused HDR image; (b) improved fused image.

that is high enough in the overlap area. Next, we can obtain the objective optimal model:

$$\begin{aligned} \min \quad & g(v_1, v_2, \dots, v_n) = \sum_{i=1}^{n-1} \sum_{j \in \Omega_i} (v_i x_{ij} - v_{i+1} x_{i+1,j})^2 \\ \text{s.t.} \quad & 0 < v_i < v_{i+1} \leq 1 \end{aligned} \quad (8)$$

Based on equation (6), let $v_n = 1$, and then use the barrier penalty function to solve the coefficients of v_1, v_2, \dots, v_{n-1} . Use equation (4) to perform fusion high dynamic projection, which contains all the information about the object under test. Thus, we can use the conventional CT reconstruction algorithm to obtain the 3D structure.

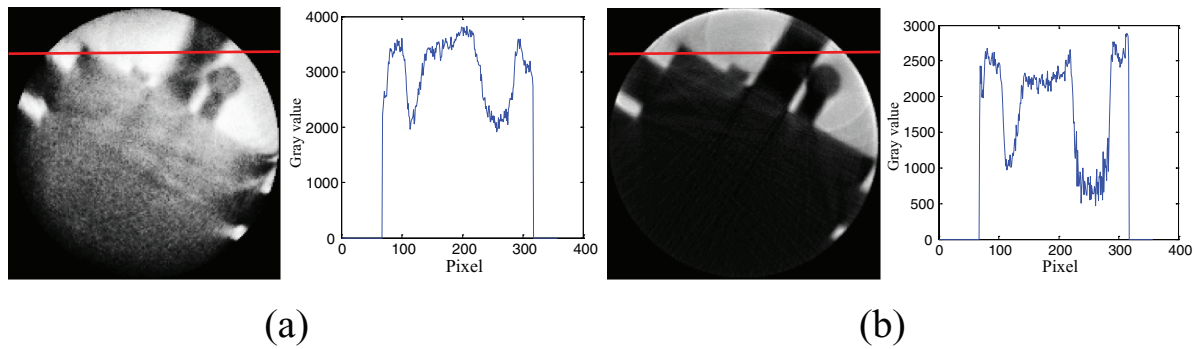


Figure 7. CT reconstruction result and grey curve at the labelled line: (a) direct fusion reconstruction and (b) logarithm demodulation fusion reconstruction.

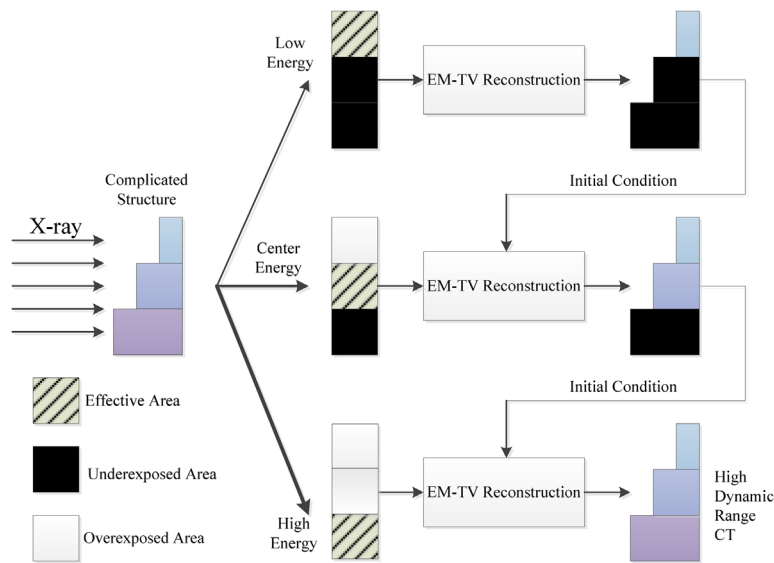


Figure 8. The functional figure of varying-energy CT reconstruction based on EM-TV.

4. High dynamic reconstruction

4.1. Directly reconstruction

For the workpiece in figure 1, use variable-energy CT scanning imaging at every projection angle to obtain an energy sequence. The energy range is from 60 kVp to 100 kVp with an energy interval of 10 kVp. For equation (8), we can also use direct fusion on the sequence, using data X_i in equation (6). Thus, we can obtain two results of direct fusion and logarithm demodulation fusion.

Figure 6 shows that the highly dynamic fusion can provide full information, and from the fusion coefficient, the logarithm fusion can achieve lower noise, as shown in figure 6(b). Compared to figure 6(a), the 3D grey curve of the logarithm fusion is smoother. To use this fusion approach to improve the CT reconstruction quality, we use the Feldkamp-Davis-Kress (FDK) algorithm to reconstruct figure 1. The results are shown in figure 7.

In figure 7(a), we combine images collected using various x-ray energies at the same projection angle and use the fusion method of the grey consistency to combine these images, which served as the input data for CT reconstruction (figure 7(a)). However, because of a weighting noise greater than 1, the

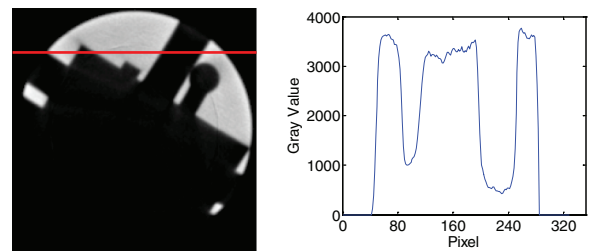


Figure 9. CT reconstruction result based on EM-TV.

Table 1. The evaluation with different CT result.

CT result	Figure 7(a)	Figure 7(a)	Figure 9
Contrast ratio	39.4%	53.6%	60.5%
Noise level	6.7%	3.8%	1.7%

CT image noise is higher and the quality is poor. To address this issue, we used negative logarithm conversion to modify the fusion coefficients to reduce the noise level. As shown in figure 7(b), the noise and contrast of CT images obtained using negative logarithm conversion are improved with respect to direct reconstruction.

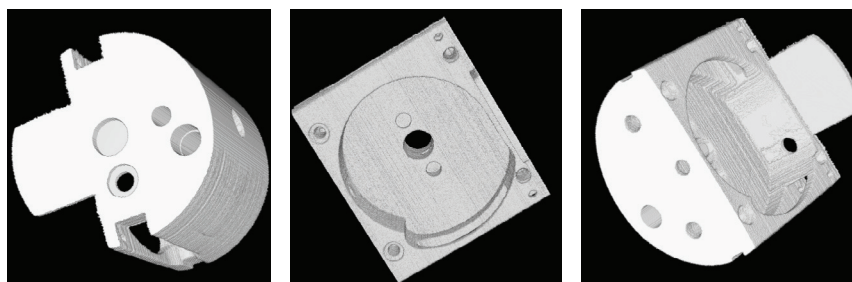


Figure 10. 3D visualizations of the real workpiece at various viewing angles.

4.2. Iteration reconstruction for varying energy

From figure 7(b), we can obtain the higher reconstruction quality; however, higher noise is still an issue because, in the lower-energy imaging, the ray dose is also lower and is thus unable to achieve a higher-quality projection. As a result, in the fusion image, this low-dose noise will influence the reconstruction quality obtained using an analytical method, such as FDK. To solve this problem, we can use an iterative method. Iterative methods with the ability to include various physical models represent a more intuitive and natural approach to image reconstruction [12]. The statistical reconstruction method, for example, models the counting statistics of detected photons by respective weighting of the measured rays. Here, we use EM (expectation maximization) statistical reconstruction to obtain the CT image [13]. In addition, to reduce noise, the minimization of total variation (TV) is used [14].

To enhance the reconstruction efficiency, we only use the fusion coefficient v_1, v_2, \dots, v_n to renew the grey, which can ensure the consistency of dynamic range. In the process of iteration, we only reconstruct the local area that corresponds to the local effective projection for the one ray energy. This result will be the initial conditions of the reconstruction of the data for the subsequent energy. In this manner, after all the data of the different energies have been processed, we obtain the full CT information. Using the system shown in figure 4, we can perform the reconstruction process shown in figure 8.

Using the above-described iterative method, we can obtain a higher-quality CT image, as shown in figure 9.

Also we give the quantitative index of contrast ratio and noise level (only in workpiece part), which is shown in table 1.

From the above result, EM-TV reconstruction has the better result of the object's detail and contrast. Also the object is only aluminium, from the grayscale consistency, the result of EM-TV (figure 9) is better than directly fusion reconstruction and logarithm demodulation fusion reconstruction. This is because EM-TV can restrain low-dose noise, improve reconstruction quality.

Because of the higher contrast ratio about figure 9, we can use threshold segmentation and the visualization toolkit to obtain 3D visualizations, as shown in figure 10. Compare the 3D visualization (figure 10) and real construct information (figure 1), the HDR-CT based on EM-TV can better represent the full construct information on the complicated workpiece.

5. Conclusion

We developed and present a novel high-dynamic-range CT imaging method and system that does not require a detector with a higher dynamic range than that of a conventional CT apparatus. This method uses multi-energy projection sequences to obtain higher-dynamic-range images via fusion, which uses the grey consistency between the adjacent images at the overlap area. Next, CT images can be obtained through logarithm conversion CT reconstruction based on the analytical reconstruction. However, the analytical method cannot effectively suppress the noise. To address this issue, we use the EM statistical reconstruction method and TV regulation to perform the reconstruction. In the process of reconstruction, we use the fusion coefficient to renew the ray image grey to the same energy level. In addition, the CT result of the one energy is the initial condition of the next one. The experimental results revealed that the CT imaging based on EM-TV has better results, with a high dynamic range and low noise level, compared to the conventional approach. Therefore this proposed approach is suitable for the imaging of complicated workpieces of widely varying thickness, which cannot be adequately imaged using conventional single-energy CT.

Acknowledgments

Project supported by the National Natural Science Foundation of China (Grant Nos. 61227003, 61301259, 61471325 and 61571404) and the Science Foundation of Shanxi Province (No. 2015021099).

References

- [1] Kalender W A 2006 X-ray computed tomography *Phy. Med. Biol.* **51** 29–43
- [2] Kim D, Park S, Lee J H, Jeong Y Y and Jon S 2007 Antibiofouling polymer-coated gold nanoparticles as a contrast agent for *in vivo* x-ray computed tomography imaging *J. Am. Chem. Soc.* **129** 7661
- [3] Philipp K and Albert W 2010 Multi-energy image stack fusion in computed tomography *Meas. Sci. Technol.* **21** 045105
- [4] Hausleiter J *et al* 2010 Image quality and radiation exposure with a low tube voltage protocol for coronary CT angiography: results of the PROTECTION II trial *JACC Cardiovas. Imaging* **3** 1113–23

- [5] Liu B, Han Y, Pan J and Chen P 2014 Multi-energy image sequence fusion based on variable energy x-ray imaging *J. X-Ray Sci. Technol.* **22** 241–51
- [6] Kalra M K *et al* 2004 Techniques and applications of automatic tube current modulation for CT *Radiology* **233** 649–57
- [7] Sisniega A, Abella M, Desco M and Vaquero J J 2014 Dual-exposure technique for extending the dynamic range of x-ray flat panel detectors *Phys. Med. Biol.* **59** 421–39
- [8] Li L, Chen Z, Cong W and Wang G 2015 Spectral CT modeling and reconstruction with hybrid detectors in dynamic-threshold-based counting and integrating modes *IEEE Trans. Med. Imaging* **34** 716–28
- [9] Persson M *et al* 2014 Energy-resolved CT imaging with a photon-counting silicon-strip detector *Phys. Med. Biol.* **59** 6709–27
- [10] Lingenfelter D J, Fessler J A, Scott C D and He Z 2013 Source detection performance prediction for a CdZnTe array *IEEE Trans. Nuc. Sci.* **60** 204–12
- [11] Chen P and Han Y 2015 Varying energy CT imaging method about complicated structural components *2015 IEEE IST* pp 1–4
- [12] Beister M, Kolditz D and Kalender W A 2012 Iterative reconstruction methods in x-ray CT *Eur. J. Med. Phys.* **28** 94–408
- [13] Green P J 1990 Bayesian reconstructions from emission tomography data using a modified EM algorithm *IEEE Trans. Med. Imaging* **9** 84–93
- [14] Yan M *et al* 2011 EM + TV based reconstruction for cone-beam CT with reduced radiation *ISVC 2011 7th Int. Symp.* (Las Vegas, NV, 26–28 September 2011) pp 1–10