The Detector System Design for the Grating-based Phase Contrast Imaging CT Prototype

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1. Introduction

X-ray radiographic absorption imaging is a standard tool in the medical diagnostics reasearch area. However, the weak absorping structures, such as biological tissue, cannot be imaged with a high sensitivity under the X-ray dosage limit. Theoretically, the interaction cross section of X-ray phase shift is usually much greater than the absorption cross section for soft tissues. Therefore, recording the X-ray phase shift can obtain substantially increased contrast. In the past years, various phase-sensitive X-ray imaging mehtods were developed, such as refraction-based method, propagation-based method, analyzer crystal-based method[1-3]. Nevertheless, the application of these methods in medical diagnostics is very slow, because a much higher quality X-ray source is required than that of conventional laboratory or hospital X-ray sources.

In 2006, a three transmission gratings phase-contrast image method with conventional X-ray tubes is reported[4]. The G0 granting creates an array of individually coherent, but mutually incoherent sources. Through the combination of gratings G1 and G2, the small angular deviation changes the locally transmitted intensity. The grating-based phase contrast imaging(GBPCI), which can achieve a large field of view with low requirement on time and spatial coherence of the light source, is considered as a potential imaging method for clinical applications. A CT prototype based on GBPCI is developed by National Synchrotron Radiation Laboratory (NSTL), and the detector system for the prototype is designed.
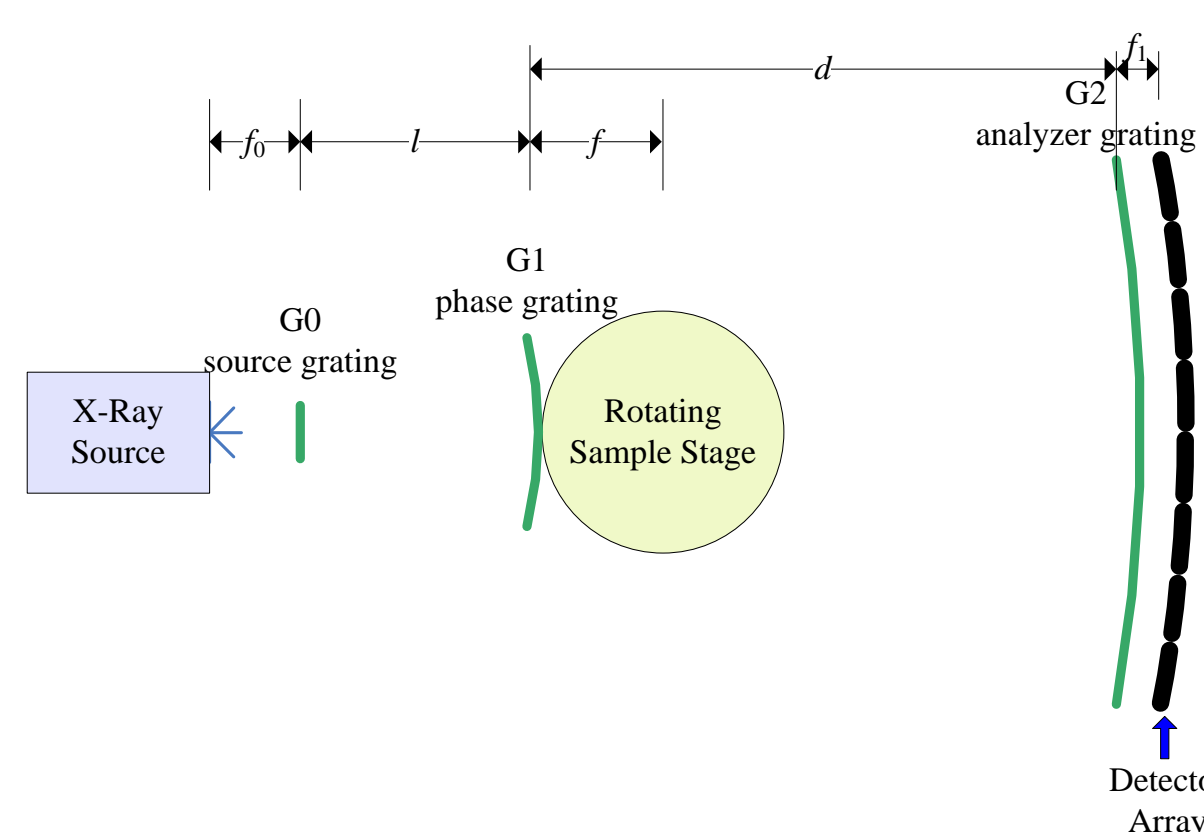

Fig.1 The schematic diagram of three grating phase-contrast image method

2. System Design

The detector system consists of 43 Hamamatsu silicon photodiode S12058(X) modules to achieve 200mm*200mm measurement field. Each S12058(X) module can provide 384(16*24) pixels, which the minimum pixel size is 0.75mm*1mm. The front-end electronics(FEE) board is direct connected to each S12058(X) module. The structure of FEE is shown in Fig.2.
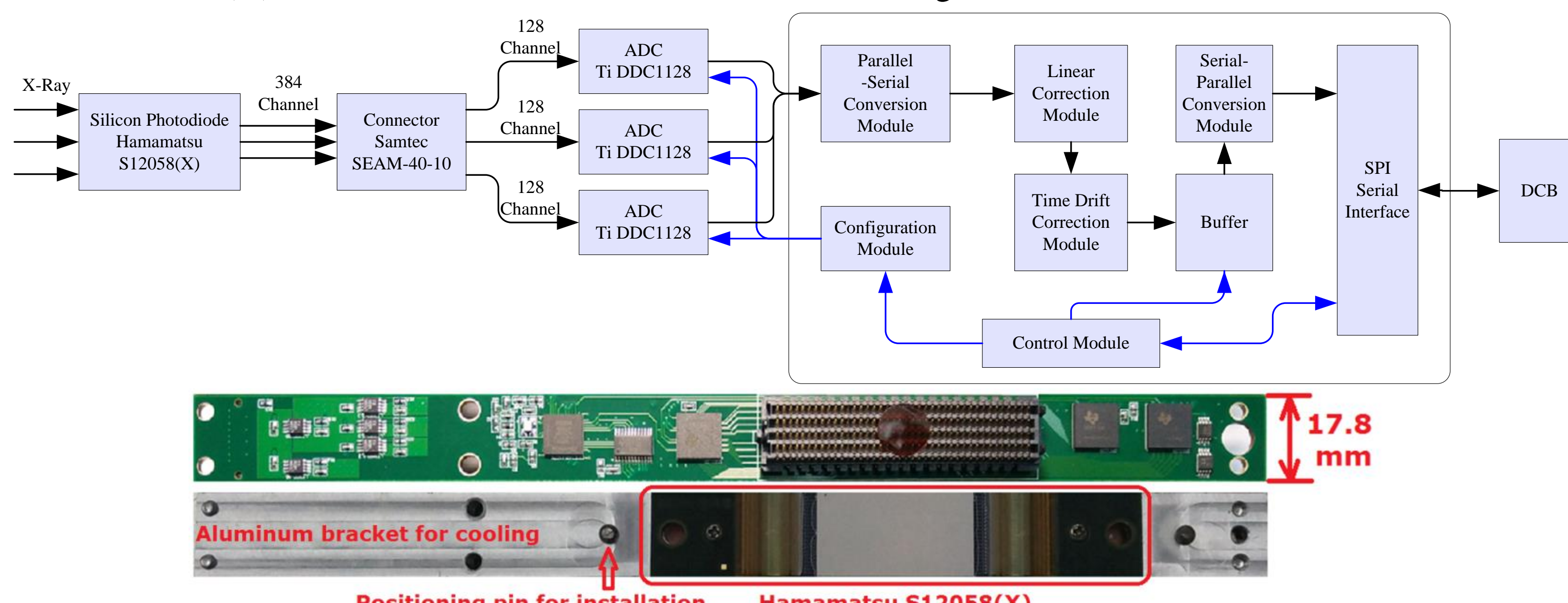

Fig.2 The schematic diagram (above) and the prototype board (below) of the FEE

The S12058(X) module receives the X-rays and generates 384 current signals whose electric charge are related to the intensity of X-rays. These current signals are transmitted to the FEE via a high density connector (SAMTEC SEAM-40-10). Three Ti DDC1128 analog-to-digital converts(ADCs) are used to complete both current-to-voltage and A/D conversion. The serial digital data transmit into the FPGA and convert to the parallel data. After the linear correction and time drift correction, the image data are sent to the data collection board(DCB) through the data buffer.

Consider to the system installation, the width size of FEE must smaller than the detector. The size of S12058(x) module is 18mm*100mm, thus the width of FEE is limited to 17.8mm. Due to the large number of pixels, the peak power consumption of each FEE can reach to 3W. An aluminum bracket with 10mm thickness is used to export the heat which generated by each FEE, and an air-cooled system is setting to cool the whole detector system.

3. Electronic Tests

A. Electronics noise

A noise test is performed to evaluate the efficient resolution of the detector system. The S12058(X) module is set to test mode, then all the pixel inputs are connected to the GND. The equicalent noise of each pixel is calculated by the Full-Width-Half-Maximum of the Gaussian distribution fitted to the output amplitude, as is shown in Fig.3(a). The average noise of the detector system is about 6.2fC. The efficient resolution of the detector system can be calculated as : N-log(σ) = 14.6bit.

B. Consistency

For the diversity of each detector module and the difference of each FEE channel, the gain uniformity test is performed. As is shown in Fig.3(b), the gain where located on the edge of the S12058(X) module is evidently lower than other pixels which caused by the edge effect. An air scan calibration method is used to correct the consistency.
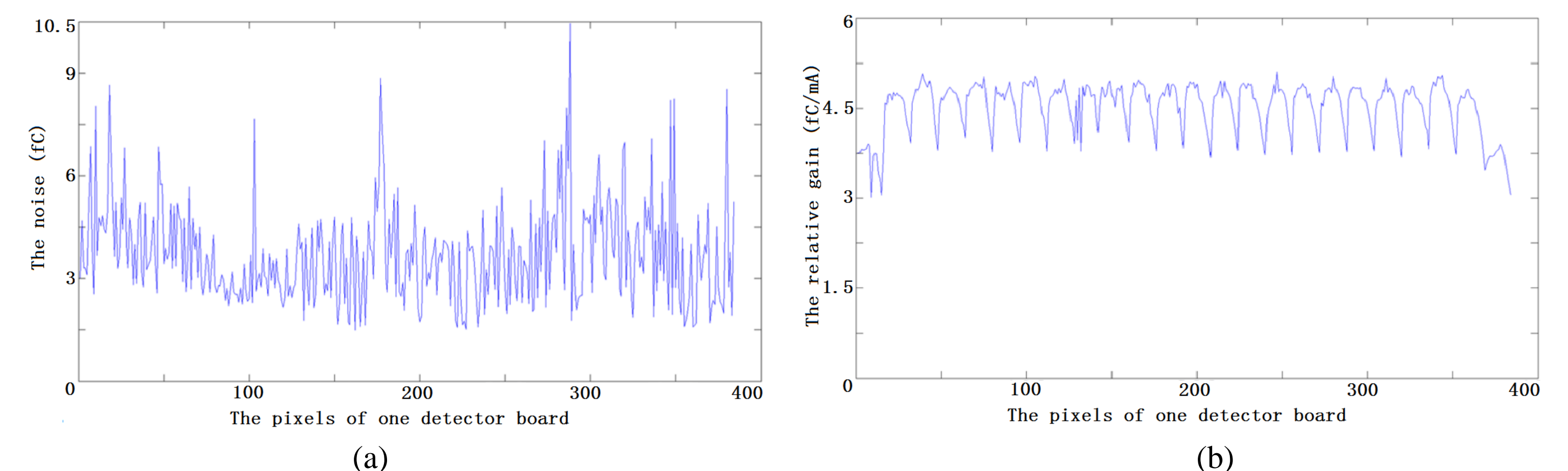

Fig.3 The equicalent noise(a) and the relative gain(b) of 384 pixels on one detector board

C. Nonlinear

The CTR2150 X-ray source is used to perform the linearity test. By stepping up the output current of the source, the incoming charge quanntity of each pixel is synchronous increased linearity. The integral nonlinearity(INL) of the detector board can be obtained with simple calculation. As is shown in Fig.4. The result shows that the detector system has a good linear response in the signal dynamic range, and the INL of 384 pixels on one detector board is less than 0.1%.
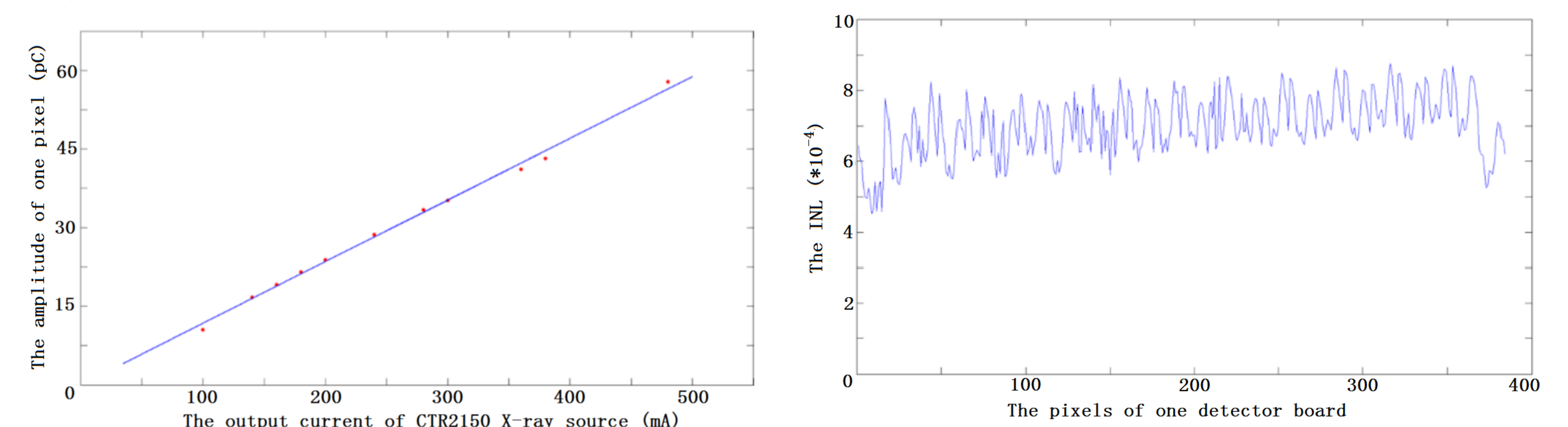

Fig.4 The linearity of one pixel response (left) and the INL of 384 pixels on one detector board

D. Spatial Resolution

A desktop experiment platform is set up to measuring the spatial resolution. The simplified detector system includes 5 detector boards to accommodate the size of the desktop platform. A lead resolution board is used as the test sample. The test result is shown in Fig.5. The spatial resolution of the detector system can be up to 0.38mm with a 17% contrast after air scan calibration.
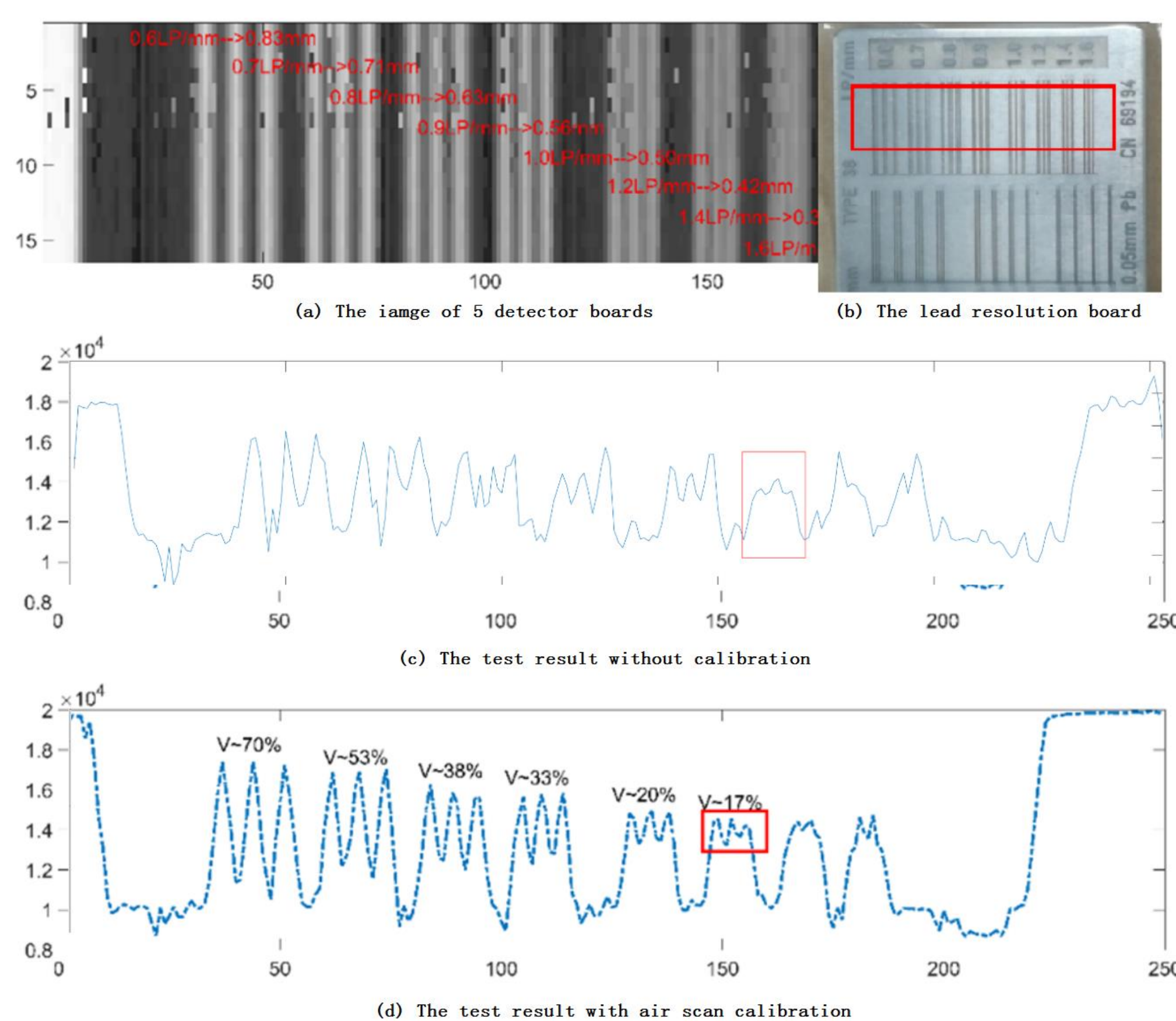

Fig.5 The spatial resolution test result of desktop experiment platform

4. Conclusions

In this paper, we present a detector system for the grating-based phase contrast imaging CT prototype. The system contains 43 detector boards including 16512 pixels. Test result shows that the average efficient resolution is about 14.6 bits. The INL of all pixels is less than 0.1%. The air scan algorithm is used to calibrate the gain non-uniformity. The spatial resolution of the system can be up to 0.38mm with a 17% contrast for the principle experiment.

References

1. A. Momose, et al. "Phase-contrast X-ray computed tomography for observing biological soft tissues," Nature Med. 2, 473–475 (1996).
2. T. J. Davis, et al. "Phase-contrast imaging of weakly absorbing materials using hard X-rays," Nature (London) 373, 595–598 (1995).
3. P. Cloetens, et al. "Holotomography: Quantitative phase tomography with micrometer resolution using hard synchrotron radiation x rays," Appl. Phys. Lett. 75, 2912–2914 (1999).
4. Pfeiffer F, et al. "Phase retrieval and differential phase-contrast imaging with low-brilliance X-ray sources." (4)2,258-261(2006).