

Stationary digital breast tomosynthesis system with a multi-beam field emission x-ray source array

Guang Yang^a, Ramya Rajaram^b, Guohua Cao^a, Shabana Sultana^b, Zhijun Liu^a, David Lalush^c,
Jianping Lu^a, Otto Zhou^{a,b,d}

^aDepartment of Physics & Astronomy, University of North Carolina, Chapel Hill, NC 27599

^bCurriculum of Applied & Materials Sciences, University of North Carolina, Chapel Hill, NC 27599

^cJoint Department of Biomedical Engineering, University of North Carolina, Chapel Hill, NC 27599
and North Carolina State University, Raleigh, NC 27695

^dLineberger Comprehensive Cancer Center, University of North Carolina, Chapel Hill, NC 27599

ABSTRACT

A stationary digital breast tomosynthesis (DBT) system using a carbon nanotube based multi-beam field emission x-ray (MBFEX) source has been designed. The purpose is to investigate the feasibility of reducing the total imaging time, simplifying the system design, and potentially improving the image quality comparing to the conventional DBT scanners. The MBFEX source consists of 25 individually programmable x-ray pixels which are evenly angular spaced covering a 48° field of view. The device acquires the projection images by electronically switching on and off the individual x-ray pixels without mechanical motion of either the x-ray source or the detector. The designs of the x-ray source and the imaging system are presented. Some preliminary results are discussed.

Keywords: digital breast tomosynthesis, carbon nanotubes, field emission

1. INTRODUCTION

Mammography is currently the most effective screening and diagnostic tool for early detection of breast cancer, and has been attributed to the recent reduction of breast cancer mortality rate¹. However, the nature of the two-dimensional mammogram makes it difficult to distinguish a cancer from overlying breast tissues, and the interpretation can be variable among radiologists. A higher rate of false-positive and false-negative test results exists because the dense tissues interfere with the identification of abnormalities associated with tumors. Digital breast tomosynthesis (DBT) is a three-dimensional imaging technique that is designed to overcome this problem. It is a limited angle tomography technique that provides reconstruction planes in the breast using projection images from a limited angular range².

Several prototype DBT scanners have been manufactured by commercial vendors including GE³, Hologic⁴ and Siemens⁵. The system designs are based on the current full-field digital mammography (FFDM) unit. A mammography x-ray tube is used to collect the projection images by moving 10-50 degrees around the object. The reported total scanning time is 7s - 40s depending on the number of views and the thickness of the breast, which is much longer than that of the regular mammography. The long imaging time causes patient motion blur which degrades image quality and make patients uncomfortable. The power of the x-ray source, gantry rotating speed and detector frame rate limit the scanning speed of the current DBT systems as illustrated by the timing diagram of the DBT system in the continues motion mode in Fig. 1. In this case X-ray radiation is only delivered within a small portion (<10%) of the total imaging time.

All the prototype DBT systems utilize the standard mammography x-ray tube with a ~300 μ m x-ray focal spot size. Due to the gantry rotation and mechanical instability, the effective focal spot size during image acquisition is larger than the static value which degrades the image resolution. Two gantry rotation modes have been developed. The GE system uses a stop-and-shoot method. The gantry makes a full stop before taking each projection image. Acceleration/deceleration may cause mechanical instability of the system⁵. The extent of this effect on the system resolution has not been fully evaluated. A continuous rotation mode is used in the Siemens and Hologic systems. The gantry keeps a constant rotation speed during the whole imaging process. In this case, the x-ray focal spot size is enlarged along the motion direction. The schematic is shown in Fig. 2. The value of the enlargement depends on the rotation speed, and the exposure time.

Ren et al.⁴ reported the x-ray focal spot moves ~1mm in a typical scan. The effect of the focal spot blurring on the system MTF was found to be not significant at this speed especially for thin specimen⁴. However it does not lead room for further reduction of the total scanning time, which will require a faster gantry rotation and a larger focal spot blurring.

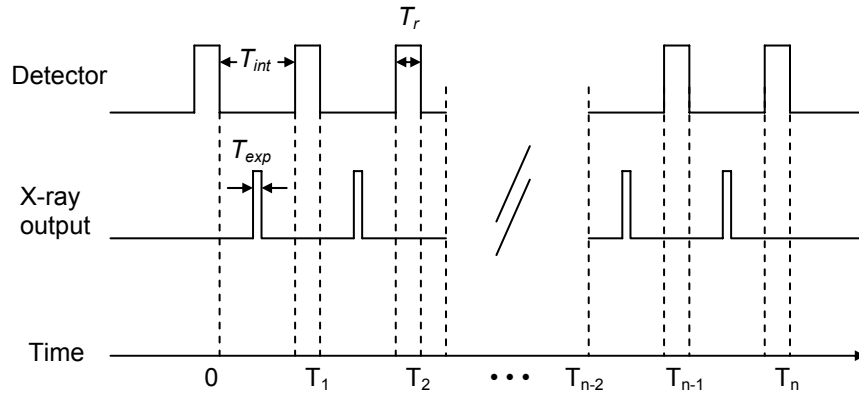


Fig. 1: Timing diagram of DBT imaging. The total imaging time is determined by $n*(T_{int}+T_r)$, where n is the view number, T_{int} is the detector integration window, and T_r is the detector readout time. T_{exp} is the exposure time for each projection image. The total exposure time is less than 10% of the total time in DBT imaging.

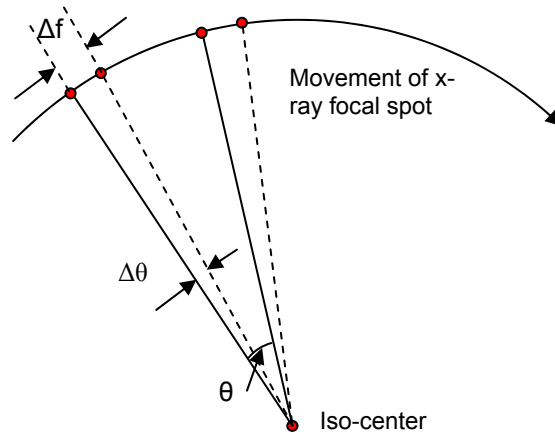


Fig. 2: Schematic of the focal spot enlargement of continuous rotation mode. The gantry rotates angle θ between two imaging positions. During the x-ray radiation period, the gantry rotates angle $\Delta\theta$ and moves Δf along the arc. Δf is the additional x-ray focal spot size induced by gantry rotation. The value is determined by the x-ray radiation period and gantry rotation speed. For a typical scan⁴, $\theta=3^\circ$, $\Delta\theta=0.12^\circ$, and $\Delta f=1\text{mm}$.

The purpose of this study is to investigate the feasibility of constructing a stationary DBT system which can increase the scanning speed, simplify the system design and potentially enhance the image quality. The enabling technology is the carbon nanotube (CNT) based multi-beam field emission x-ray (MBFEX) recently developed in our lab⁶⁻⁸. For demonstration of principle a stationary DBT scanner was designed and constructed at UNC using a spatially distributed MBFEX source with 25 individually controlled x-ray pixels. The device, called Argus, acquires the projection images without mechanical motion of the sources or the detector. In this paper the basic designs of the source and the system are described and the some preliminary results are discussed.

2. SYSTEM DESCRIPTION

2.1 System geometry

A picture of the assembled Argus system is shown in Fig. 3(a). The system is composed of a CNT MBFEX source, a flat panel x-ray detector (Varian Medical System, Paxscan 2520), a control unit, and a computer work station. The system geometry follows the typical values for the regular mammography devices, as shown in Fig. 3(b). The distance between the center of the phantom and the x-ray source is 64.5cm. The source to detector distance is 69.6cm, which leaves a 2.5cm air gap for a normal 5-cm breast phantom. The x-ray pixels are arranged linearly to reduce the system complexity, with even-angular distribution and a 2-degree increment. The total angular coverage of the x-ray source is 48 degrees. In such a design, the distance between the nearest x-ray focal spots varies from 2.5cm to 2.7cm, and the total span of the x-ray source array is 57.5cm.

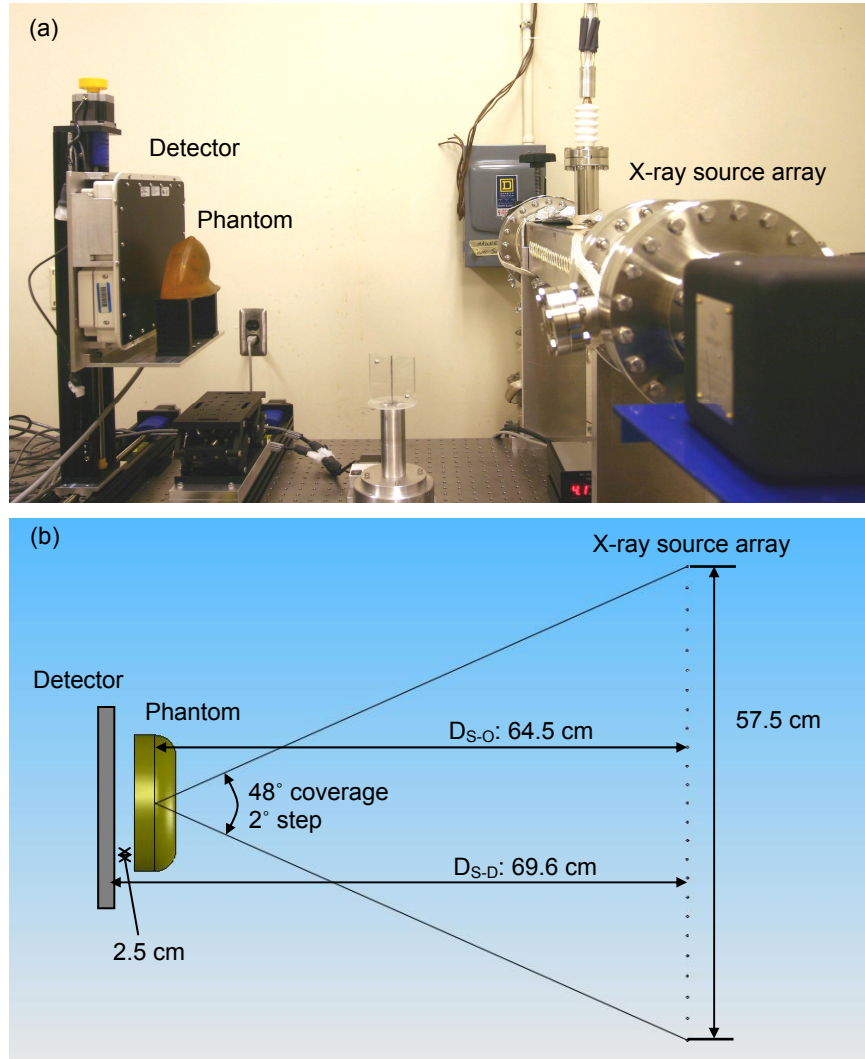


Fig. 3: (a) Picture of the Argus system. The x-ray source array, breast phantom, and detector are shown. (b) Geometry of the Argus system. The 25 x-ray focal spot positions are represented by 25 points.

2.2 X-ray source array

The key component of the Argus system is the field emission x-ray source array. The construction of the 25 x-ray pixels is identical. The CAD drawing of one pixel is shown in Fig. 4. The central beams of all the 25 x-ray sources intersect at a

common iso-center, which is also the center of the phantom. Each x-ray pixel consists of one CNT cathode, one gate electrode, two focusing electrodes, and one anode. The design of each x-ray pixel is similar to that of the field emission micro-focus x-ray source we have published previously⁹. The cathode is grounded, and all other electrodes are maintained at constant voltages during imaging acquisition. The gate voltage determines the x-ray tube current. Below a threshold, there is no current, and the current increases exponentially with gate voltage once over the threshold. The electron trajectory is controlled by the voltages applied to the two focusing electrodes. The anode is made of molybdenum. The x-ray source array is housed in a customized chamber. A 30 μm molybdenum film serves as the x-ray window as well as the filter. The molybdenum anode and filter combination is commonly used for breast imaging¹⁰. Up to 40keV high voltage can be applied on the anode. The targeted performance for the source design is that each x-ray pixel can provide 10mA peak current at 200 μm x200 μm effective focal spot size.

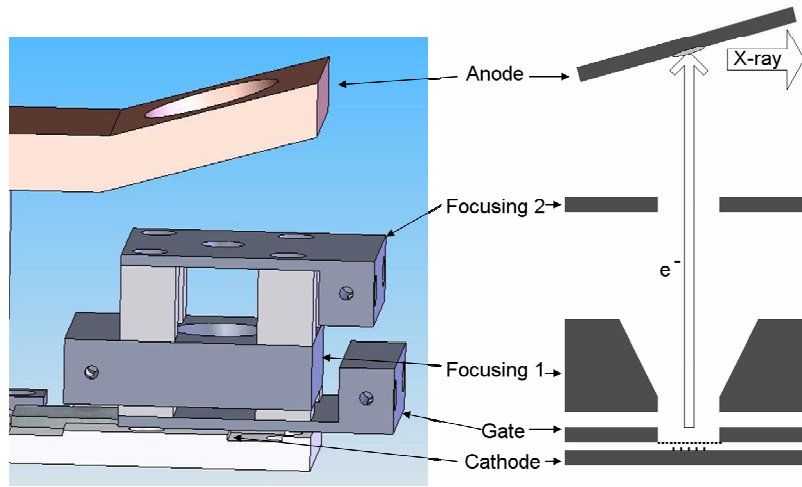


Fig. 4: CAD drawing of one x-ray pixel in the array. Each pixel consists of a CNT cathode, a gate electrode, two focusing electrodes, and one molybdenum anode.

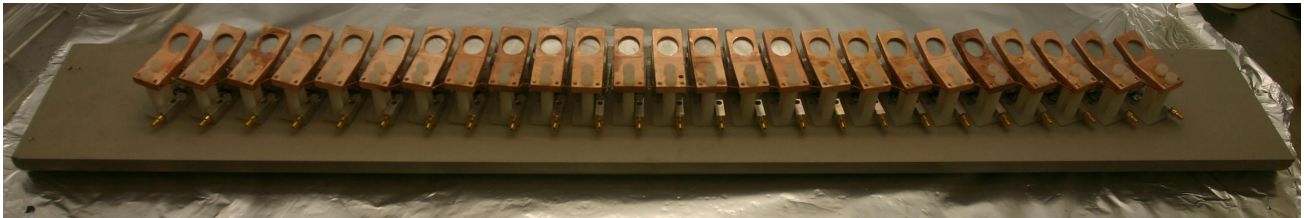


Fig. 5: An assembled MBFEX source array with 25 individually controllable x-ray pixels which are tilted towards the iso-center.

A control unit based on MOSFET (metal-oxide-semiconductor field-effect transistor) is built to switch the x-ray sources. The basic design of the electronics switching system is similar to the one used in our first demo MBFEX source¹¹. The diagram is shown in Fig. 6. The drains (D), gates (G), and sources (S) of the 25 MOSFET's are connected to the 25 cathodes, 25 TTL trigger signals (provided by the computer board), and a common ground, respectively. When the TTL trigger signal is at low state, the conduction channel between the source and the drain is closed. This causes the CNT cathode potential floating relatively to the common ground, and no electron is emitted from the CNT cathode and no x-ray radiation is generated. When the TTL trigger signal is at high state, the cathode is grounded because of the opened conduction channel. As a result, electrons are extracted by the electrical field between gate electrode and cathode, and x-ray radiation is produced. The delay time (between switching of the TTL signal and the conduction channel) of the MOSFET is 35-45 ns, which is sufficient considering the tens of milliseconds x-ray exposure period. The 25 x-ray pixels can be switched individually at any given time during the imaging acquisition, which provides great flexibility. Variable resistors are built-in for compensation of the variations in the individual cathode performance.

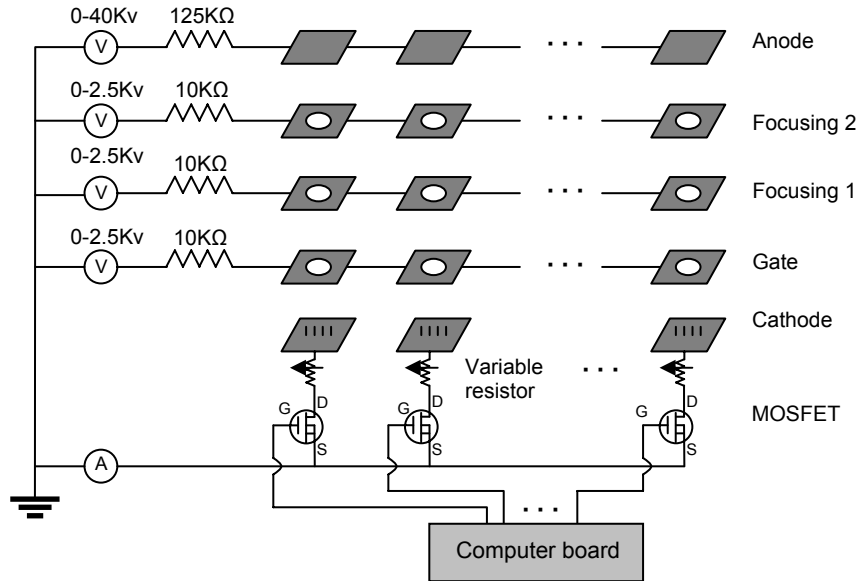


Fig. 6: Schematic of the control unit. Through the control unit, each x-ray source can be switched individually by the computer.

2.3 Detector and image acquisition software

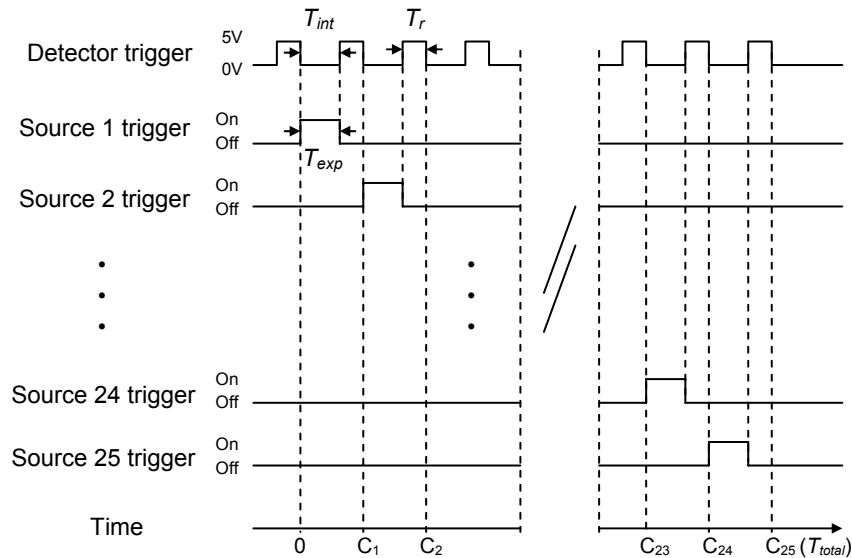


Fig. 7: Timing diagram of Argus system. The detector readout and 25 x-ray sources are triggered by TTL signals provided by computer board. The x-ray sources are turned on sequentially, and readouts are triggered after each radiation.

A flat panel detector (Varian Medical System, Paxscan 2520) is used for imaging acquisition. The field of view is 19.5cm by 24.4cm, which ensures a full image of the breast. With a $127\mu\text{m}$ pixel pitch, the total array size is 1536 by 1920. The detector can run under non-binning mode and 2 by 2 binning mode. In the user synchronization mode, the rising edge of a continuous TTL signal triggers the readout of the detector, as illustrated in Fig. 7. The imaging time is determined by the integration window T_{int} and the detector readout time T_r . T_{int} is controllable through triggering signal. X-ray radiation is delivered within the integration window T_{int} , and the radiation period is denoted as T_{exp} . The readout time T_r depends on the acquisition mode. For the normal and 2 by 2 binning modes, the readout time is 128ms and 32ms,

respectively. In the Argus system, the 25 projection images are taken sequentially. The total imaging time can be calculated by $25*(T_{int} + T_r)$.

An imaging acquisition software is developed using LabView. The software controls a computer board to trigger both the x-ray pixels and the detector, and also acquires the x-ray images from the detector. The exposure time and tube current of each x-ray source can be adjusted through the software. The software is designed for sequential imaging protocol and can be modified for other imaging sequences.

2.4 Reconstruction program

To reconstruct the slice images, an iterative ordered-subset convex (OSC) algorithm¹² based on a maximum-likelihood (ML) model is used. The reconstruction technique applies a shearing method to convert all projection images to a common frame of reference; then uses a pre-computed cone-beam model to project and back-project in the common frame. To reduce the computational load, non-cubic voxels are reconstructed. This technique has been verified on both simulated data and breast phantom images measured from a field emission x-ray source array with limited number of pixels¹³.

2.5 Comparison with the prototype systems

The designed system performance of the UNC Argus DBT system is listed in Table 1, along with the specifications of 3 prototypes from commercial vendors based on the information from recent scientific publications³⁻⁵. The potential advantages of this system include the followings. 1: The focal spot size of Argus system is 200 μ m while the values of other systems are 300 μ m or larger. 2: Stationary design gives less gantry vibration by eliminating the mechanical movement. 3: The exposure time matches the detector integration window. In other words, the utilization of the imaging time is the most efficient. 4: The targeted total scan time (8.8s in binning mode and 11.2s in full-resolution mode, 25 viewing angles) is shorter which can be further reduced by increasing the x-ray tube current which requires relaxing of the focal spot size.

Table 1: Specification of Argus and that compares with other prototype DBT systems.

	UNC: Argus	GE: Senographe 2000D	Siemens: Mammomat Novation	Hologic: Selenia
X-ray kVp, mA	25-35kVp, 10mA	25-30kVp, ~130mA	~28kVp, ~180mA	24-39kVp, ~100mA
Focal spot size	200 μ m	300 μ m	300 μ m + blur*	300 μ m + blur*
Target/filter	Mo/Mo	Mo/Mo, Rh/Rh	W/Rh	(Mo, W)/(Rh, Al)
Angle coverage	48 degrees	50 degrees	50 degrees	30 degrees
View numbers	25	11	25/49	11
Gantry motion	Stationary	Step and shoot	Continuous rotation	Continuous rotation
Flat-panel Detector	A-silicon	Cs:I a-silicon	Direct converter a-selenium	Direct converter a-selenium
Detector size	19.5 x 24.4 cm pixel pitch: 127 μ m	18.00 x 23.04 cm pixel pitch: 100 μ m	23.9 x 30.5 cm pixel pitch: 85 μ m	24 x 29 cm pixel pitch: 70 μ m (140 μ m for DBT)
Readout time	0.128s/0.032s	0.3s	0.6s/0.3s	0.6s
Integration time	0.32s	0.4s	0.2s	1.0s
Exposure time	0.32s	~0.1s	~0.03s	0.073s
Total scan time **	11.2s for 25 views	7s for 11 views	20s/39.2s for 25/49 views	18s for 11 views
Reconstruction method	ordered subsets convex (maximum likelihood)	ML-EM	FBP: filtered back projection	FBP: filtered back projection

*: Additional focal spot blur due to the gantry movement during exposure.

** : Total scan time = (view number) x (cycle time); cycle time = (readout time) + (integration time).

3. PRELIMINARY SYSTEM CALIBRATION

3.1 Spectrum measurement

The energy spectrum of the x-ray source was measured at 28keV using a Si-pin photodiode detector from Amptek (XR-100CR and MCA-8000A). The spectrum is consistent when measured at different locations within the field of view, and from different x-ray pixels. The experimentally measured energy spectrum is shown in Fig. 8, which agrees well with typical Mo/Mo x-ray spectrum¹⁰. Two molybdenum characteristic peaks, one at 17.5keV and the other at 19.6keV, are well recognized.

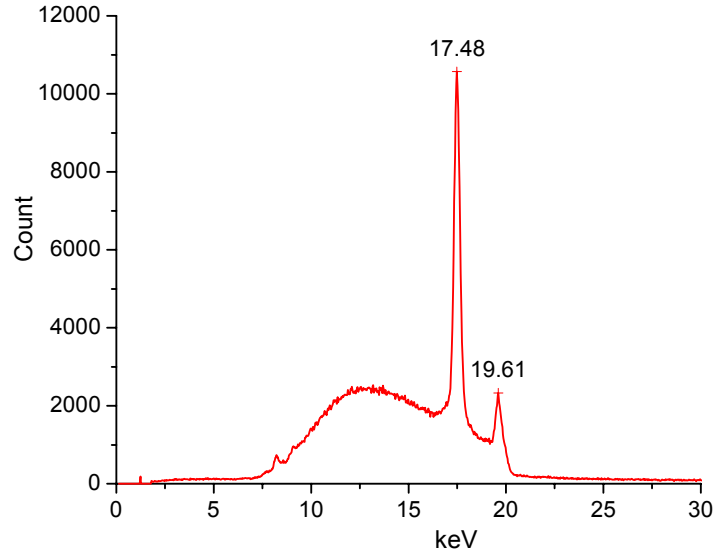


Fig. 8: The experimentally measured energy spectrum of the UNC Argus system at 28keV.

3.2 X-ray tube current

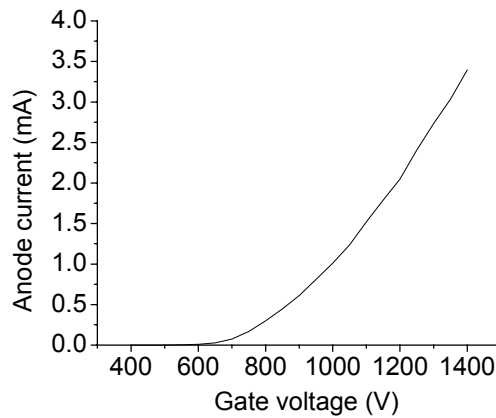


Fig. 9: A typical curve of anode current as a function of gate voltage. The threshold value for this x-ray source is about 650V.

The emission current from the CNT cathode depends on the electrical field between the gate and the cathode following the known Fowler-Nordheim equation¹⁴. In this particular x-ray source, 72% of the total current passes through the gate electrode and reaches the anode to produce x-ray radiation (also denoted as anode current). Fig. 9 shows the typical anode current versus gate voltage data measured from one pixel in the MBFEX source. Due to the voltage limitation of

the electronic devices, presently the maximum gate voltage can be applied is 1500V, which limits the anode current to ~4mA, below the targeted value of 10mA. This can be overcome in the future by changing the design and/or optimizing the CNT cathode (when measured in a separate setup, the cathodes fabricated under same conditions can produce over 10mA consistently at higher gate voltages).

So far nine of the pixels in the MBFEX source have been characterized. Due to variation in the CNT cathodes, the gate voltages needed to obtain the same current are different. As a reference, the voltages vary from 925V to 1465V for 1mA tube current (Table 2). The voltage difference can be compensated by the variable resistors in the control circuit. With the improvement of fabrication technique and cathode quality control, the variation can be reduced in the future. The current stability was determined by measuring the current of 100 pulsed x-rays at constant voltage. The standard deviation of the current is less than 0.03mA for all pixels tested.

Table 2: Gate voltages required to obtain 1mA tube current and the tube current fluctuation at the same gate voltage.

X-ray source #	Gate voltage (V)	Standard deviation of current (mA)
1	1230	0.02
2	925	0.01
3	1230	0.02
4	1015	0.01
5	1300	0.03
6	1070	0.01
7	1160	0.01
8	1465	0.02
9	1030	0.01

3.3 Focal spot measurement

The designed x-ray focal spot size is 200x200 μ m for all 25 x-ray sources. The actual values were measured following the European standard EN12543-5¹⁵. A customized cross wire phantom made of 1mm tungsten wire was fabricated to measure the focal spot size along two orthogonal directions simultaneously. The phantom was placed close to the x-ray source to obtain the large magnification factor. The voltages applied to the two focusing electrodes were first varied to optimize the focal spot size. It was found that the optimal focal spot size is achieved when the two focusing electrodes are at 500V and 1600V, respectively. A typical projection image of the cross phantom is shown in Fig. 10 (a). The line profiles are shown in Fig. 10 (b).

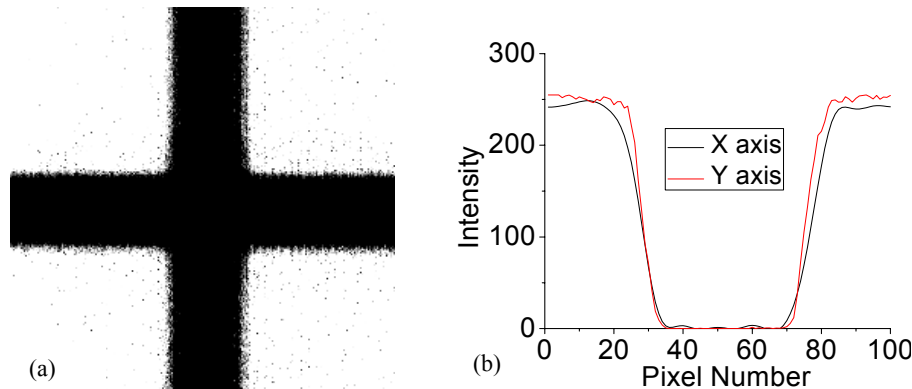


Fig. 10: (a) A typical x-ray projection image of the cross wire phantom. (b) Line profiles of the two wires. X axis is the direction of the x-ray source array, and Y axis is perpendicular to it.

Focal spot sizes of 9 x-ray pixels have been measured. As listed in Table 3, the results agree well with the designed specification 0.20 x 0.2mm. The x-ray sources have isotropic focal spot with average value of 0.19mm. Measurements from different x-ray sources are also consistent.

Table 3: Focal spot size measurement of 9 x-ray pixels. The uncertainty of the measurement is 0.02mm.

X-ray source #	F_x : parallel to x-ray source array	F_y : perpendicular to x-ray source array
1	0.20mm	0.20mm
2	0.20mm	0.17mm
3	0.18mm	0.19mm
4	0.19mm	0.19mm
5	0.20mm	0.19mm
6	0.19mm	0.17mm
7	0.18mm	0.17mm
8	0.19mm	0.19mm
9	0.18mm	0.19mm

3.4 Geometry calibration

The tomosynthesis reconstruction requires precise system geometry parameters. We applied an analytic method based on identification of ellipse parameters¹⁶ for the geometry calibration, which was first established for cone-beam CT calibration. A phantom with two point objects with known distance was machined. The geometry parameters of the 25 x-ray sources were calibrated individually. Six projection images of the phantom (60-degree rotation in-between) were acquired for each pixel. The traces of the two balls form two ellipses on the detector plane. The parameters, including the source-detector distance and x-ray source offset values on the detector plane, can be further calculated based on these elliptical curves. The source-detector distance is calculated to be 69.3cm with 2mm uncertainty. The distances between the x-ray sources are also calculated. The results agree with the design values within 1mm uncertainty.

4. CONCLUSION AND DISCUSSION

A stationary digital breast tomosynthesis system with a CNT based distributed x-ray source has been designed and assembled, with comparable geometry to the conventional mammography devices. The system has been fully assembled. The control unit, imaging software, and the calibration procedure have been developed. The preliminary test performed so far shows that the system functions as designed. Detailed system calibration and tomosynthesis imaging test will be performed in the near future.

ACKNOWLEDGEMENTS

This work was partially supported by grants from NIH-NCI (U54CA119343) and the UNC Lineberger Comprehensive Cancer Center. We acknowledge helpful discussions with Elodia Cole and Etta Pisano.

REFERENCES

- ¹ Bassett, L.W. and et al., [*Diagnosis of Diseases of the Breast*] 2 ed., Elsevier Saunders, Philadelphia, PA (2005)
- ² Dobbins, J.T. III and Godfrey, D.J., "*Digital x-ray tomosynthesis: current state of the art and clinical potential*". *Phys. Med. Biol.* **48**: R65-R106 (2003)
- ³ Wu, T. and et al., "*Tomographic mammography using a limited number of low-dose cone-beam projection images*". *Med. Phys.*, **30**(3), 365-380 (2003)
- ⁴ Ren, B. and et al., "*Design and performance of the prototype full field breast tomosynthesis system with selenium based flat panel detector*". *Proc. SPIE*, **5745**, 550-561 (2005)

- ⁵ Bissonnette, M. and et al., "*Digital breast tomosynthesis using an amorphous selenium flat panel detector*". Proc. SPIE, **5745**, 529 (2005)
- ⁶ Zhang, J. and et al., "*Stationary scanning x-ray source based on carbon nanotube field emitters*". Appl. Phys. Lett., **86**, 184104 (2005)
- ⁷ Zhang, J. and et al., "*Multiplexing radiography based on carbon nanotube field emission X-ray technology*". Proc. SPIE, **6510**(65100W) (2007)
- ⁸ Zhang, J. and et al., "*Multiplexing radiography using a carbon nanotube based x-ray source*". Appl. Phys. Lett., **89**, 064106 (2006)
- ⁹ Liu, Z. and et al., "*Carbon nanotube based microfocus field emission x-ray source for microcomputed tomography*". Appl. Phys. Lett., **89**, 103111 (2006)
- ¹⁰ Beutel, J. and et al., [*Handbook of Medical Imaging*], SPIE Press, Bellingham, WA (2000)
- ¹¹ Zhang, J. and et al., "*A stationary scanning x-ray source based on carbon nanotube field emitters*". Appl. Phys. Lett., **86**, 184104 (2005)
- ¹² Kamphuis, C. and Beekman, F.J. "*Accelerated iterative transmission CT using an ordered subsets convex algorithm*". IEEE Trans. Med. Im., **17**, 1101-05 (1998)
- ¹³ Lalush, D.S., and et al., "*Tomosynthesis reconstruction from multi-beam x-ray sources*". Proceedings of 2006 IEEE International Symposium on Biomedical Imaging, 1180-1183 (2006)
- ¹⁴ Gomer, R., [*Field Emission and Field Ionization*], Harvard University Press, Cambridge, MA (1961)
- ¹⁵ *European Committee for Standardization: EN 12543-5, E. Brussels.* (1999)
- ¹⁶ Noo, F., and et al., "*Analytic method based on identification of ellipse parameters for scanner calibration in cone-beam tomography*". Phys. Med. Biol., **45**, 3489-3508 (2000)