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14. ABSTRACT Using a proof-of-concept bench-top set-up we recently demonstrated the feasibility of stationary digital breast tomosynthesis (s-DBT) utilizing a carbon nanotube (CNT) based distributed x-ray source array. The device generates the projection views by electronically activating the multiple x-ray beams. The approach has the potential to provide a fast scanning speed without sacrificing the spatial resolution by eliminating any motion blurring from the tube motion. The purpose of this study is to evaluate the feasibility of improving the spatial resolution and increasing the scanning speed by replacing the conventional scanning mammography tube with a stationary x-ray source array. We have successfully evaluated the features of CNT based distributed x-ray source array for improving the spatial resolution and increasing the scanning speed of the conventional DBT scanner.					
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Introduction

Digital breast tomosynthesis (DBT) is a three-dimensional (3D) imaging technique that provides the reconstruction of an arbitrary set of planes in the breast from limited-angle series of projection images as the x-ray tube moves. Compared with traditional two-dimensional (2D) mammography, 3D tomosynthesis imaging methods have the potential to improve conspicuity of structures by removing the visual clutter associated with overlying anatomy [1,2,3]. The DBT reconstruction methods make it possible that breast cancer can be detected at smaller size and earlier stage, thereby reducing the number of women who die each year from breast cancer [4]. However, a typical 3D tomosynthesis scan takes about 20 seconds to more than 1 minute (depends on how many projection views are acquired). In order to speed up the scan time, binning mode was implemented by manufacturers to average neighboring pixels at the expenses of sacrificing image quality and resolution [3]. In addition, to reduce the motion blurring in DBT scanner with a single moving x-ray source requires a large increase in the x-ray tube power. However, the peak power of the high power x-ray tubes is approaching its physical limit [5].

The effective way to reduce scanning time and the motion blurring is to utilize a spatially distributed x-ray source which can radiate x-ray radiations from different viewing angles without mechanical motion. This has not been possible until the development of the multi-beam field emission x-ray (MBFEX) source by our team [6]. Contrary to conventional thermionic cathodes, carbon nanotube (CNT) based MBFEX source has several intrinsic advantages: high temporal resolution and capabilities for spatial and temporal modulation [7].

Our **hypothesis** is to develop the next generation DBT scanner with significantly improved system performance at potentially reduced dose and cost. The **rationale** for this hypothesis arises from the following factors. All current commercial DBT scanners have several intrinsic limitations: 1) the source rotation leads to long scanning time [8]; and 2) the slow motion of the source leads to motion blurring and system instability that limited the spatial resolution [9]. We propose to develop a novel stationary DBT (s-DBT) scanner to mitigate the above limitations.

Body

Task 1. Design and construct a fully functional full-field s-DBT scanner with the specially designed high-power MBFEX source (Months 1-17)

The aim is to design a scanner with a faster scanning time, better spatial resolution, and the capability to perform dual energy imaging at a reasonable scanning time. In order to perform a realistic comparison of performance, the configuration of the scanner is similar to the Hologic Selenia Dimensions scanner. The following specific works have been carried out in year one:

1.1 Define the specifications of the s-DBT scanner and the MBFEX source. (Months 1-6)

This specified aim has been successfully completed.

The goal of this work is to define the specifications of the proposed scanner and the x-ray source. The scanner has been developed to generate a typical tomosynthesis sequence with enough projection images. The array of X-ray sources are fixed in a straight line that is parallel to the stationary detector plane. This is different from the geometry currently used by the commercial prototype scanners, where the x-ray source mounted on a C-arm rotates along an arc. The configuration of the s-DBT scanner, in terms of angular range, number of views, scanning time etc., is tabulated in Table 1.

The x-ray anodes are tilted towards the center of the object (defined as the isocenter). The centers of the focal points remain on a straight line that is parallel to the detector plane. The 31 x-ray beams in the x-ray tube span a distance of 30cm from end-to-end. At a source-object distance of 70 cm, it provides 30° coverage, with an even 1° angular

spacing between each beam (the linear spacing between the x-ray pixels varies to provide the even angular spacing). The x-ray beams are collimated to cover the whole flat panel detector. A flat panel detector is used for imaging acquisition.

Table1: Specifications of the s-DBT scanner	
Number of views	31
Angular range	30 degrees
Scanning time	Down to 4sec
Source-to-detector distance	70cm
Object-to-detector distance	2.5cm
x-ray tube current	Up to 29mA
Cooling	Active anode cooling
Beam geometry	Cone beam
x-ray anode voltage	Up to 50kV
X-ray anode/filtering	Tungsten/Al

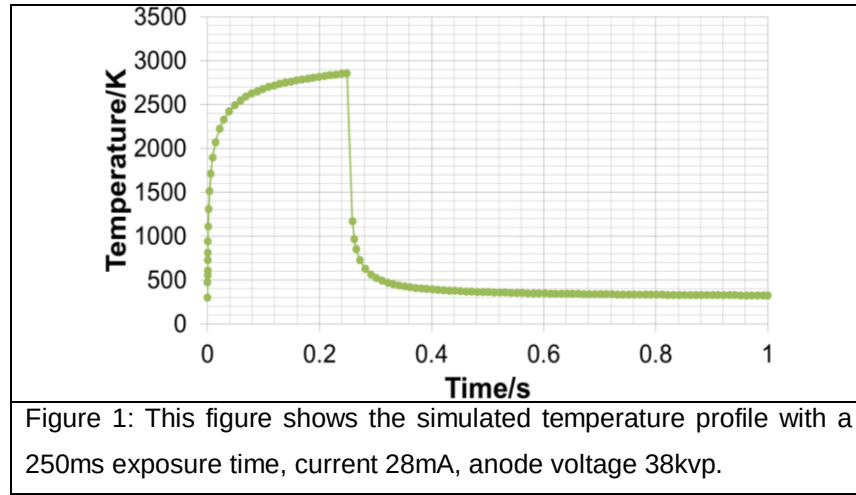
1.2 Manufacture the MBFEX source based on the specifications defined in 1.1. (Months 7-15)

An x-ray chamber and MBFEX source have been made for this proposed work. They are based on the specifications defined in aim 1.1. The optimized design of this x-ray tube is based on the performances of the current s-DBT scanner in our lab and some research works that have been performed. The specific works have been performed are listed as following:

1.2.1 X-ray tube anode heat load simulation will be performed to determine the operating power of the x-ray tube. We will perform electron optics simulation to evaluate the three focusing optics designs. The goal is to select the design that gives the required focal spot size with less design complications. (Months 7-8)

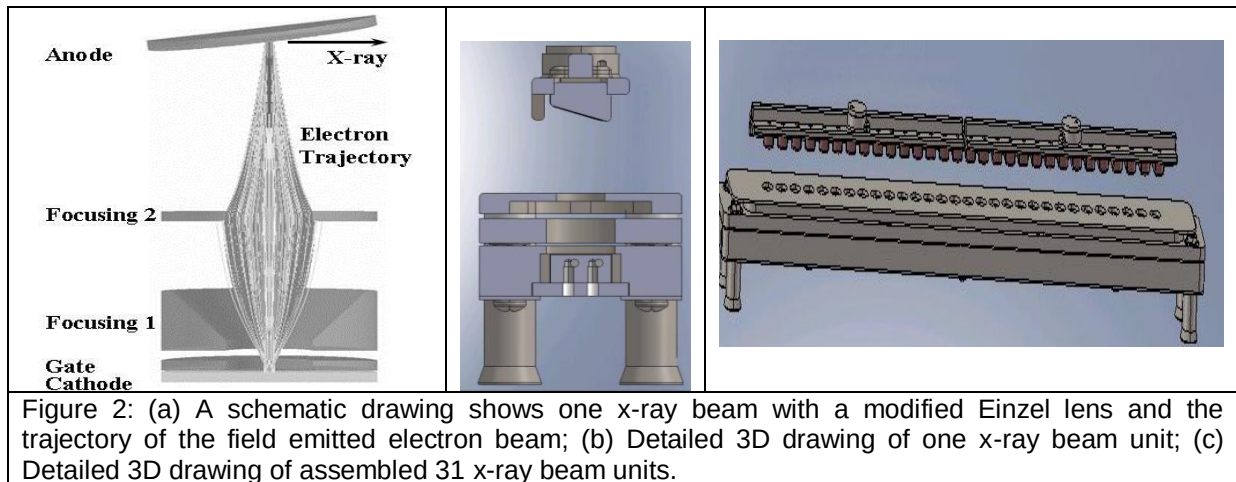
The main source of thermal energy generation in a field emission x-ray tube, as well as a thermionic x-ray tube, results from bombardment of the high kinetic energy electrons on the metal anode. About 99% of the kinetic energy of the electron is converted to heat on the anode. A conventional rotating anode mammography x-ray tube operates at 100-200mA tube current at ~30kVp anode voltage, resulting in a peak power of 3-6kW. The rotating anode design is used to distribute the thermal load over a large area. In the case of the MBFEX source designed for the proposed s-DBT scanner, the anode is an extended metallic structure. The thermal energy is distributed over the full length and the anode block has sufficient heat capacity to absorb the thermal energy generated during the scan. The energy input of 100mAs x 40kV = 4kJ leads to a very small temperature rise compared with temperature rise in conventional x-ray tubes. The peak power of the MBFEX source is only ~1kW. This power has to be removed from the tube to allow a scan of every 10mins. However there is a concern whether the individual focus spot will be heated up too close to the melting temperature of the target material for the time duration the electron beam is on. For a 100mAs exposure with 15 projection views, each focus spot is exposed for 250ms and 125ms respectively for the 4s and 2s scanning protocols.

Finite element simulation is used to study the thermal energy dissipation with a tungsten anode. The Transient Thermal Module on ANSYS 12.1.0 is used to simulate the heat load on the anode. The simulation is setup on a $60mm \times 30mm \times 5mm$ Tungsten anode with a focal spot size of $2.5mm \times 0.5mm$ (effective FSS $0.68mm \times 0.5mm$) and fully temperature dependent thermal parameters. Anode high voltage is 38kVp, current is 28mA. The power density distribution at the focal spot is implemented with Gaussian distribution. Temperature profile is simulated with a 250ms exposure time. The simulation result is shown in Figure 1. Under the 250ms exposure, the anode can generally stand the heat generated. From the maximum temperature profile, we can see the anode temperature drops down quickly so that the anode is not affected much by the heat generated by accelerated electrons.

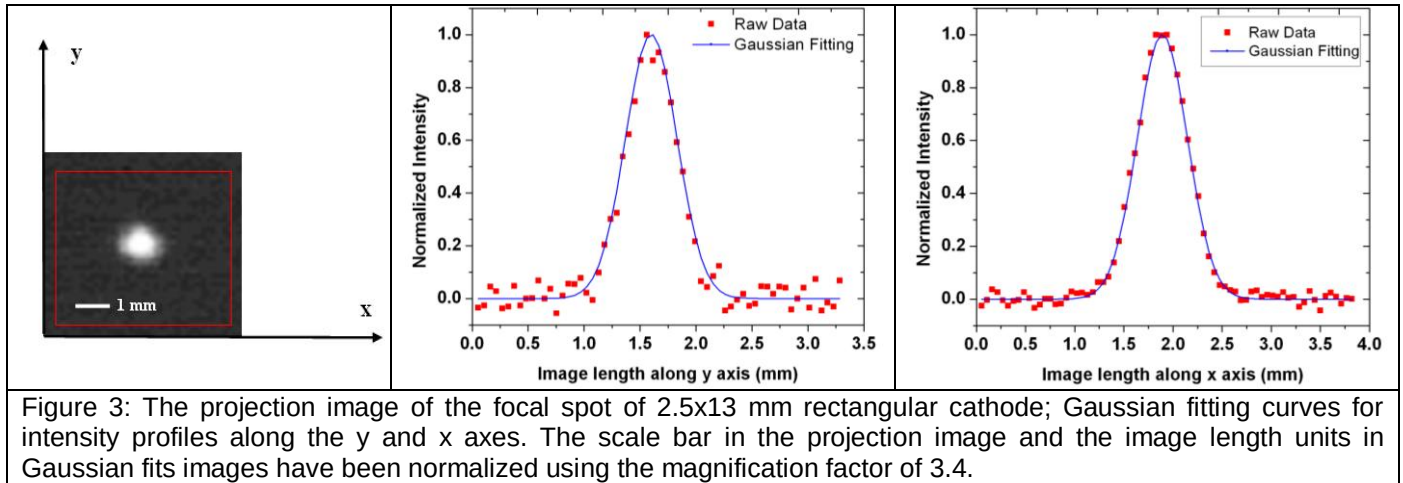


The aim of electron optics design is to select the appropriate focusing optics for the proposed x-ray source. We have performed computer simulations using a commercial software package (VectorField) to evaluate the modified Einzel lens with three active focusing electrode designs. The goal is to select the design that gives the required focal spot size with less design complications. Figure 2 (a) shows the schematic drawing of the x-ray beam with a modified Einzel lens and the trajectory of the field emitted beam. Figures 2 (b-c) show the detailed 3D drawings of an individual x-ray beam unit and 31 assembled beam units. These 31 assembled beam units will be put into the constructed x-ray chamber.

Each x-ray generating unit comprises a CNT field emission cathode, a gate electrode to extract the electrons, and two modified Einzel-type electrostatic lenses to focus the field emitted electrons to a small area on a Tungsten anode. The anode, focusing electrodes 1, and 2, and the cathode will be electrically connected to the corresponding electrodes on the adjacent x-ray generating units. They will be connected to respectively the following 4 power supplies: Spellman SR70PN6 (70KV, 85mA), Stanford Research PS350 (2.5KV, 10mA), and Keithley 248 (5KV, 5mA), Glassman ER03R100 (3KV, 100mA).



The effective FSS of the x-ray source has been measured using a gold-platinum pin-hole phantom. The diameter of the pinhole is $d=100\mu\text{m}$ with 500 μm length and 12° opening angle. The obtained FSS results are shown in Figure 3. The FSS measurement will be continued using different methods regarding the IEC standard.

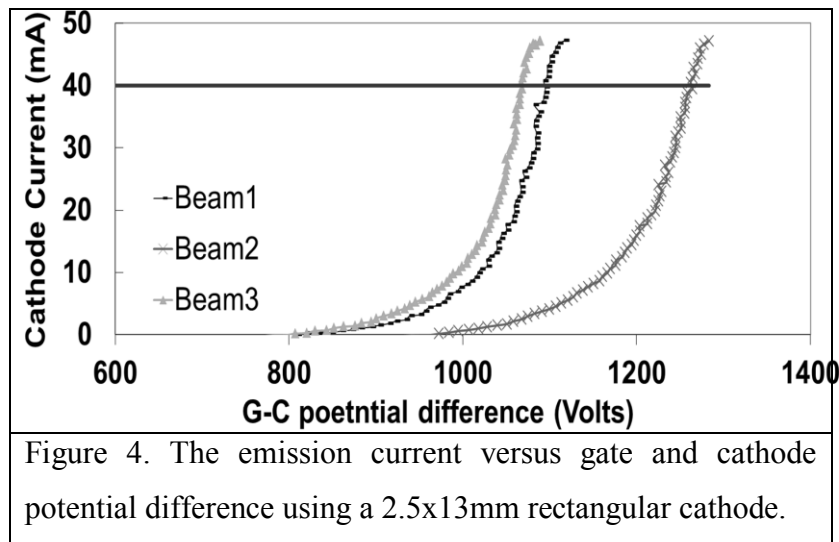


1.2.3 X-ray tube fabrication: The MBFEX source will be made at UNC. The tube manufacturing and process work will be done at XinRay, Siemens is one of its two parent companies. The tube will have an open design with a flange that can be opened and re-closed. Therefore, the internal structure of x-ray source including the anode and the cathode array can be easily changed. (Months 9-15)

The x-ray source has been modified for the specifications of this project. The electron source arrays are under fabrication at XinRay's facility in RTP, NC. The tube manufacturing and process work will be done at Siemens Medical Solution's, one of its parent companies, facilities in Germany. The x-ray tube is purchased using a different grant.

Key Research Accomplishments

The cathode current measured recently in this study readily reached 40mA using 2.5x13 mm cathodes, as shown in Figure 4. The gate and cathode potential difference among three beams can be easily compensated by using our home made adjustable resistor box. Figure 5 shows the lifetime measurement results using 2.5x13 mm rectangular cathodes. The experiment was performed at peak cathode current 40mA, pulse width 250ms. The pulse waveform in the subfigure of Figure 5 shows the pulse shape of cathode current, the pulse width is 250ms. The current value is $3 \times 40\text{mA}$, it is because of the way we measured the current using the Textronics current probe. As we can see, the cathode shows very little degradation after 8000 minutes measurement time, the total x-ray on time is 424 minutes. Since the measurement data we acquired is too large to be plotted out using OriginLab software, we plotted three sections of the whole data: beginning, middle and last sections. The trend-line is used to connect these three sections. For an imaging protocol using 100mAs per tomosynthesis scan with 15 x-ray beams, the dose from each x-ray beam is $\sim 6.7\text{mAs}$, which translates to 10mAs per cathode using the measured transmission rate of 67% (67% of the cathode current reached the anode). Assuming the tube is operating at the cathode current of 40mA (a stable emission at this level has been readily obtained at 1262 V applied cathode voltage), the required exposure time is 0.25s per cathode. The 424 mins electron beam-on time translates to an x-ray source array lifetime of at least 101760 scans. For a busy mammography clinic with ~ 60 patients per day, the results means that MBFEX source can be operated for at least 1696 working days at this output level. If we consider 250 working days per year, this MBFEX source can last for ~ 7 years. During measurement the gate voltage essentially remained the same, indicating the CNT cathode can be operated for far longer time than this.



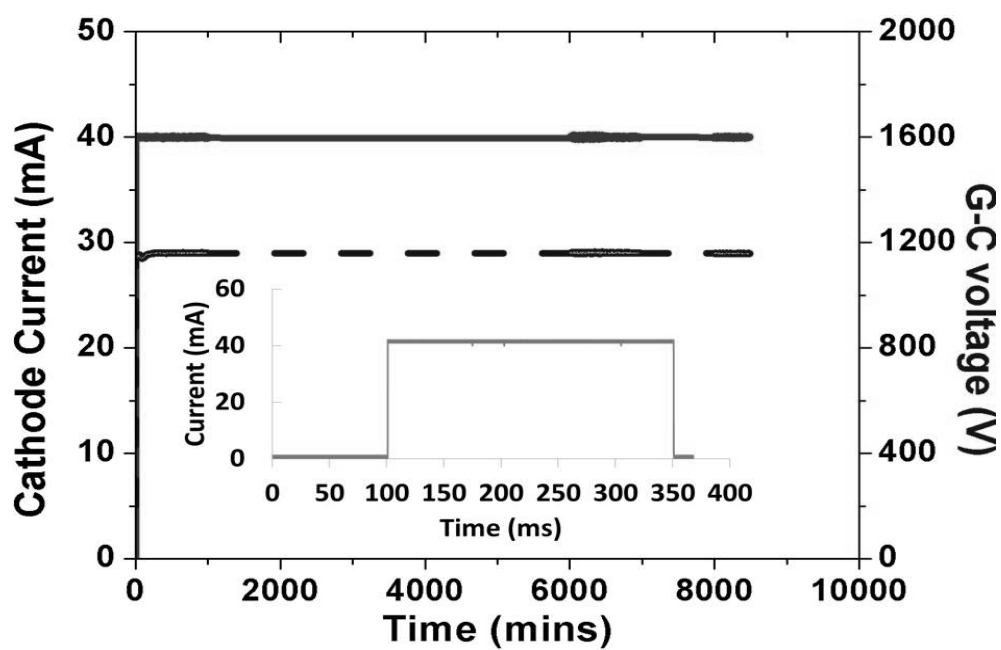


Figure 5: lifetime measurement performed at peak cathode current 40mA, pulse width 250ms. G-C voltage (~1200v) is the potential difference between gate electrode and cathode. The pulse waveform in the subfigure shows the pulse shape of cathode current.

Reportable Outcomes

Works for year one resulted in some research publications which include:

Journal Articles:

1. A Stationary Distributed X-ray Source Array for High Resolution Digital Breast Tomosynthesis, Medical Physics, **X. Qian**, G. Yang, Calderon-Colon, J. Shan, A. Tucker, JP. Lu, O. Zhou. **(in process)**
2. Design and Feasibility Studies of a Stationary Digital Breast Tomosynthesis System, G. Yang, **X. Qian**, T. Phan, F. Sprenger, Nuclear Instruments and Methods in Physics Research A, (2010), doi:10.1016/j.nima.2010.11.147

Presentations:

3. **X. Qian**, JP. Lu, O. Zhou, etc.: A Spatially Distributed X-Ray Source Array for High Resolution Digital Breast Tomosynthesis, SU-C-301-3, Joint AAPM/COMP Meeting, Vancouver, Canada, August 2011.

Conclusion

Year 1: (May 2010 – May 2011): All aims were achieved.

- Completion of the s-DBT system specification and design
- Completion of the MBFEX specification and design
- Beginning construction of the MBFEX source and x-ray tube

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Appendices

See the following attachments.



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Design and feasibility studies of a stationary digital breast tomosynthesis system

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ABSTRACT

Studies have shown that digital breast tomosynthesis (DBT) can improve breast cancer diagnosis by reconstructing 3D images. However, DBT scanners based on rotation gantry prolong the imaging time and reduce spatial resolution due to motion comparing with the regular two-view mammography. To obtain three dimension reconstruction images and maintain the high image quality of conventional mammography, we proposed a prototype stationary digital breast tomosynthesis system (s-DBT). The proposed s-DBT system acquires projection images without mechanical movement. The core component of the s-DBT system is a specially designed spatially distributed multi-beam X-ray tube based on the carbon nanotube field emission X-ray technology. The multi-beam X-ray source array enables collection of all projection images from different viewing angles without mechanical motion. Preliminary results show the s-DBT system can achieve a scan time comparable with the regular two-view mammography, and improve the spatial resolution comparing with rotating gantry DBT.

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1. Introduction

Digital breast tomosynthesis (DBT) [1] is a three dimensional imaging technique which reconstructs slice images using projections from a certain angular range. DBT can potentially improve the tumor conspicuity over conventional mammography by reducing tissue overlapping [2]. Current DBT scanners are all based on single X-ray source full-field digital mammography (FFDM) systems. To acquire projection images from different angular positions, the X-ray gantry rotates during the scanning. The total scan time is much longer than exposure time of conventional mammography. The prolonged imaging time introduces patient motion blur, and the effective X-ray focal spot size is enlarged to 1 mm [3] or larger due to the X-ray source movement during exposure. Both effects degrade the image quality, which limits further improvement on breast cancer diagnosis.

In this study, we proposed and demonstrated a stationary DBT prototype scanner, which can overcome some of the limitations of the current DBT scanners. The design and preliminary test result are reported. Potential clinical applications are discussed.

2. Material and methods

2.1. Stationary digital breast tomosynthesis

Current DBT scanners require the physical movement of the X-ray tube, which is the limiting factor in the scanning speed. To eliminate the gantry motion, we proposed the concept of stationary digital breast tomosynthesis or s-DBT scanner [4,5]. As illustrated in Fig. 1, the s-DBT system is composed of a stationary X-ray source array, control electronics for the X-ray source, a flat panel detector, a detector controller, and a computer workstation. The X-ray source array replaces the rotation gantry. The X-ray source array consists of spatially distributed X-ray beams which are individually addressable. The switching of each X-ray beam is achieved through the control electronics and the computer workstation. The X-ray radiation is synchronized with the detector readout with accuracy better than 1 ms.

During a tomosynthesis imaging, projection X-ray images from different angles are acquired sequentially by switching the X-ray beams one by one. By eliminating the gantry movement, the s-DBT scanner can reduce the scan time and decrease the patient motion blur. The X-ray focal spot enlargement induced by rotating motion is also eliminated thanks to the stationary design. The image quality can be improved comparing with the DBT scanners with rotation gantry.

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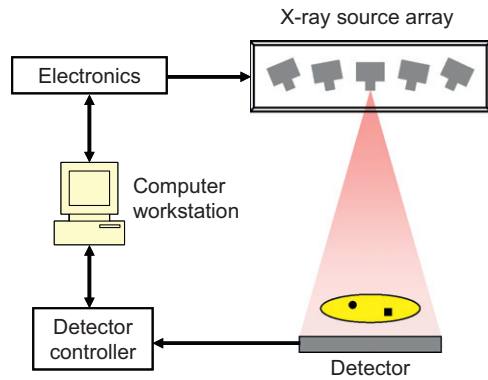


Fig. 1. Schematic of s-DBT system.

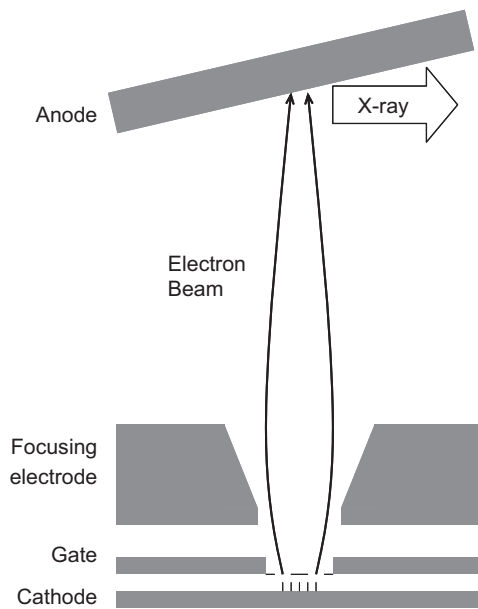


Fig. 2. Schematic of field emission X-ray source.

2.2. Field emission X-ray source array based on carbon nanotubes

A novel field emission X-ray source technology based on carbon nanotube is applied in the proposed s-DBT system. The design of the cathode structure and electron beam optics to generate a single X-ray beam is similar to the microfocus X-ray source developed for a micro-CT system [6]. The schematic is shown in Fig. 2. The X-ray source is composed of a carbon nanotube cathode, a gate electrode, a focusing unit, and an anode. Carbon nanotube field emitters are deposited at a predefined small area on the substrate. The emission current is proportional to the cathode size and follows the Fowler–Nordheim equation [7]. Electrons are emitted under a bias gate voltage

$$I = aV^2 \exp(-b\phi^{3/2}/V)$$

where a and b are constants, and ϕ is the work function of the field emitter.

By using modified asymmetrical Einzel-type focusing lenses, the trajectory of the electrons can be well controlled. A demagnification from the cathode size to focal spot size up to 5 has been demonstrated [6].

Comparing with conventional thermionic cathodes, the novel X-ray source has several intrinsic advantages, including high temporal resolution and the capabilities of spatial and temporal modulations. The feasibility of field emission X-ray source array based on carbon

nanotubes has been studied [8]. The tomosynthesis imaging has been demonstrated with the prototype system with five X-ray beams.

3. Results

3.1. Bench-top prototype s-DBT scanner

A bench-top s-DBT scanner has been developed [4]. Fig. 3 shows the system schematic with CAD drawing of the X-ray source array. This prototype system consists of a linear X-ray source array with 25 individual beams, a flat panel detector (model: PaxScan 2520, Varian Medical Systems, Salt Lake City, Utah USA), and control electronics. The source-to-detector distance is 69.6 cm, which is comparable with conventional mammography scanners. To fully utilize the X-ray radiation, the 25 X-ray beams are tilted towards the system iso-center, which coincides with the center of the breast phantom. The angular coverage of the 25 X-ray beams is 48° with 2° separation between projections. A LabView program has been developed to control the scanner and acquire projection images. The s-DBT scanner has been fully characterized. System calibration result was reported earlier [9], including spectrum measurement, focal spot size, and geometry calibration.

Due to the performance variation, the required gate voltages are different to generate same currents from the 25 beams. To compensate the gate voltages, 25 variable resistors were integrated in the control circuit. To obtain constant X-ray output across all 25 beams, I – V curves of the 25 cathodes were measured and resistor values were adjusted prior to imaging.

A Stereotactic needle biopsy tissue equivalent breast phantom (Model 013, Computerized Imaging Reference Systems, Inc., Norfolk, Virginia, USA) was imaged on the prototype system. Anode voltage was 28 kVp, filter/target combination was Mo/Mo, and the total exposure level was 100 mAs. The images were reconstructed to 60 slices (slice thickness of 1 mm) of size 1200×700 ($0.1 \times 0.1 \text{ mm}^2$) using Modified Ordered Subsets Convex (MOSC) algorithm. The slices shown in Fig. 4 are 3 mm apart and clearly show the various masses getting focused at different depths.

The spatial distribution of the X-ray beams is different from the s-DBT systems based on rotation gantry. Several other reconstruction algorithms, such as filtered back projection (FBP) and matrix inversion tomosynthesis (MITS), have also been modified to take into account the linear source geometry in the s-DBT system. Comparison between different algorithms has been reported [10].

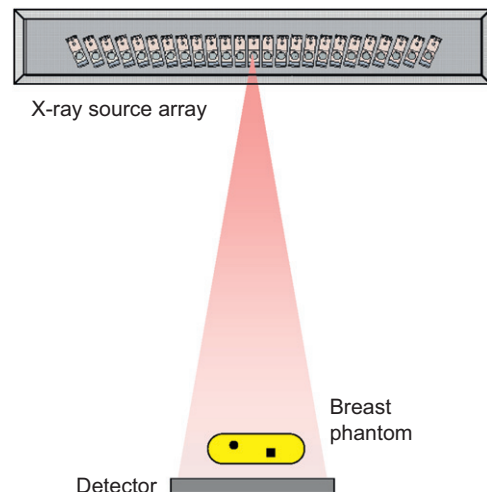


Fig. 3. Schematic of the bench-top s-DBT system and X-ray source array CAD drawing.

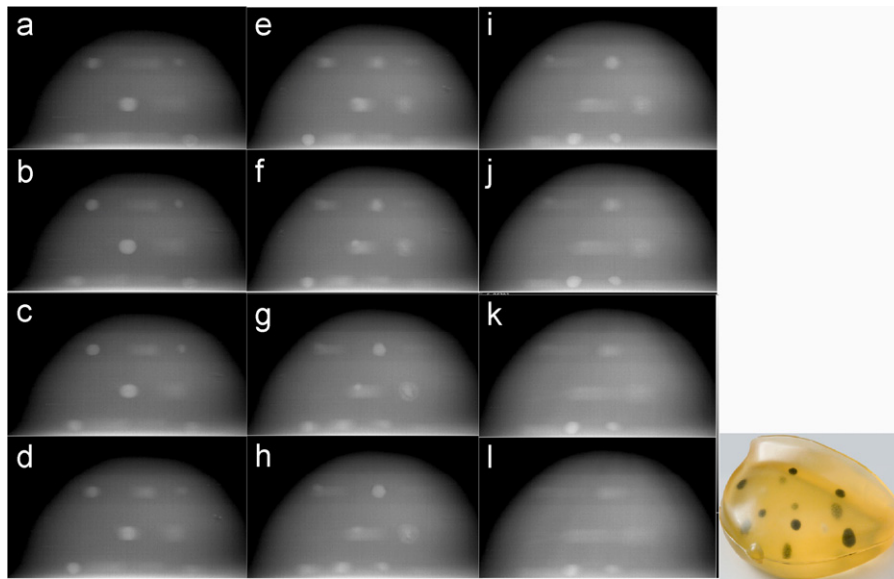


Fig. 4. Reconstructed slice images of a breast phantom. Slices are 3 mm apart. The picture of the phantom shows the masses embedded inside.

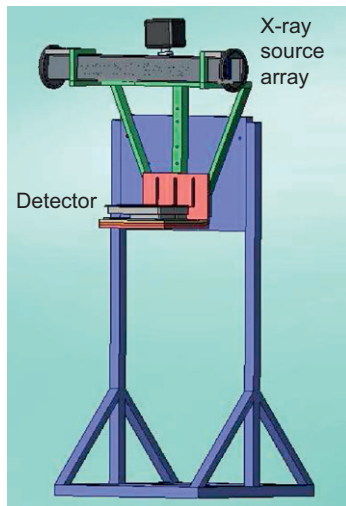


Fig. 5. CAD drawing of the s-DBT scanner for clinical application.

3.2. Prototype s-DBT scanner for clinical application

To further investigate the clinical potential of the s-DBT, a second prototype scanner is under construction (Fig. 5). The flat panel detector is placed horizontally and the X-ray source array is above the detector. The X-ray source array and detector can be rotated and fixed at desired angles for MLO view imaging. Additional collimators and shielding are also designed to meet FDA regulations [11]. The same control electronic system will be used on the new s-DBT scanner.

The other objective of the new scanner is to reduce the scan time. Because the field emission current is proportional to the cathode emission area, larger cathodes are fabricated for the new scanner to obtain higher X-ray tube current. Fig. 6 shows the emission current measurement result of 19 cathodes with $2 \times 8 \text{ mm}^2$ emission area. Two cathodes have lower performance (broken lines in the diagram) and these cathodes will be screened out. The other 17 cathodes (solid lines in the diagram) can generate 30 mA current at electric field less than $7 \text{ V}/\mu\text{m}$. As the transmission rate (the ratio of anode current to cathode current) is usually 70% [9], the anode current can reach 20 mA. The new s-DBT scanner

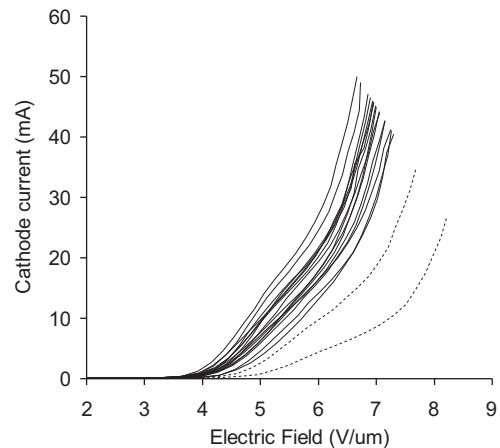


Fig. 6. Field emission measurement of 19 cathodes with $2 \times 8 \text{ mm}^2$ emission area.

can reduce the scan time to 6 s per scan based on 100 mAs dose level, (most full field digital mammography and DBT scans require less than 100 mAs radiation dose [12]).

4. Conclusion and discussion

A bench-top s-DBT scanner has been successfully built and characterized. The preliminary result shows the s-DBT design can potentially reduce the scan time and improve the image quality. With further optimization of electron beam focusing structure, it is possible to increase anode current to 60 mA, this will reduce the scan time further to be around 2 s, comparable with scan time of typical 2-view mammography. The second prototype s-DBT scanner has been designed for clinical studies. The system is capable for both CC and MLO scans. It only takes 10 s to change the projection mode. The system will be characterized and compared with other DBT systems.

The s-DBT system enables some advanced imaging techniques for breast imaging. The spatially distributed X-ray sources are ideal for dual energy tomosynthesis imaging. With conventional DBT systems, dual energy imaging requires two scans. The time delay between the two scans may introduce noise and reduce image quality. With the s-DBT systems, the X-ray sources can be tuned at two energy levels. This can

be achieved by either applying different anode voltages or filtering X-ray beams with different materials. The proposed schematic is that X-ray beams at odd numbers are tuned at low energy and X-ray beams at even numbers are tuned at high energy. In such a design, a single scan can produce both 3D and energy dispersive images. The low and high energy tomosynthesis images will be first processed separately, then combined to resolve dual energy information.

The s-DBT system can also take advantage of the multiplexing technique to further reduce the scan time [13]. By simultaneously acquiring multiple projection images, the total scan time can be reduced up to a factor of $N/2$, where N is the number of X-ray beams. Because of the reduction of the scan time, the s-DBT system makes quasi-monochromatic imaging feasible. This could potentially reduce the patient dose and improve image quality. These techniques will be studied with the new s-DBT system.

Acknowledgements

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A Spatially Distributed X-ray Source Array for High Resolution Digital Breast Tomosynthesis

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Purpose: Using a proof-of-concept bench-top set-up we recently demonstrated the feasibility of stationary digital breast tomosynthesis (s-DBT) utilizing a carbon nanotube (CNT) based distributed x-ray source array. The device generates the projection views by electronically activating the multiple x-ray beams. The approach has the potential to provide a fast scanning speed without sacrificing the spatial resolution by eliminating any motion blurring from the tube motion. The purpose of this study is to evaluate the feasibility of improving the spatial resolution and increasing the scanning speed of the Hologic Selenia Dimension DBT scanner by replacing the scanning mammography tube with a stationary x-ray source array.

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Methods: Electrostatic simulations were carried out to design the electron focusing optics to achieve the desired focal spot size. Finite element analysis was performed for both the anode and the extraction electrodes to determine the maximum x-ray power can be generated at the targeted focal spot size. Electron field emission current from the carbon nanotube cathode was measured to assess the maximum stable tube current that can be generated at the desired focal spot size. A control electronic system is developed to scan and to regulate the imaging dose from each beam by compensating the driving voltages and by modulating the exposure time from each beam. A spatially distributed multi-beam field emission x-ray (MBFEX) source array is designed for an imaging geometry similar to that of the Hologic Selenia Dimension tomosynthesis scanner. The tube can be operated in either 15 or 31 view mode with the targeted scanning speed of 4 seconds in both cases.

Results: A new electron focusing optics has been developed to achieve the desired x-ray focal spot size (FSS) 0.7x0.6mm. The measured results showed the targeted FSS has been achieved. Thermal simulations using Ansys software has been successfully finished to determine the maximum anode heat load. The field emission measurement of the CNT cathodes confirmed that maximum stable tube current can be generated at the desired focal spot size. The stability of the cathode has been tested by performing a lifetime run for over 8000 min running at 250ms pulse width, 5s period at 40mA cathode current. As an initial test, 15 projection images have been acquired using 3.3mAs per projection at 28kvp. A projection image acquired using central beam is shown in this article.

Conclusions: We have successfully evaluated the features of CNT based distributed x-ray source array for improving the spatial resolution and increasing the scanning speed of the Hologic Selenia Dimension DBT scanner. The experimentally measured FSS of 2.5x13mm rectangular cathodes is 0.7x0.6 mm, which is favorable compared to that of Hologic DBT scanner with motion blurring. FSS is correlated to output power and heat load on anode. Given certain size of focal spot on anode, the heat load on anode increases accordingly with the increase of output exposure level. The energy input of 100mAs x 40kV = 4kJ leads to a very small
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temperature rise. Our thermal simulation indicated that the anode block has sufficient heat capacity to absorb the thermal energy generated during the scan. Furthermore, the individual focus spot heated up during the exposure time is below the melting temperature of the target material Tungsten. The dissipation of heat is fast enough to allow a typical scan of every 10 minutes. We have experimentally demonstrated that tube current of 27mA can be achieved, therefore scanning time is as fast as 4s for the total imaging dose of 100mAs. This value is comparable with the reported scanning times of Hologic DBT scanners. The CNT stability test shows that this MBFEX source can last for a few years for a busy mammography clinic. These preliminary results demonstrate the feasibility of the proposed s-DBT scanner. The technology has the potential to increase the resolution and reduce the scanning time for DBT.

Keywords: x-ray, carbon nanotube, field emission, digital breast tomosynthesis

1. INTRODUCTION

Mammography plays a central part in early detection of breast cancers because it can show changes in the breast up to several years before a patient or physician can feel them [1]. Breast cancer mortality rate has been reduced due to uses of mammography [2]. While mammography is the best screening tool for breast cancer available today, it does not detect all breast cancers and is known to have a high false positive and false negative rate [3]. A variety of new breast imaging techniques are currently under development that can potentially contribute to the early detection of breast cancer and improve the accuracy in distinguishing non-cancerous breast conditions from breast cancers [1, 4-13]. Digital breast tomosynthesis (DBT), which is a limited angle computed tomography technique, has shown significant early promises in addressing these imaging limitations [14]. Tomosynthesis consists of taking x-ray images at multiple angles about a stationary compressed breast. The images are reconstructed into a 3-D dataset, which can be viewed in thin slices with high in-plane resolution that do not suffer from tissue overlap confusion. It has the potential to improve the effectiveness of early breast cancer screening using a similar dose as the common 2-D mammography [15]. Several DBT systems from commercial vendors are currently under clinical trials [5, 10, 13, 16, 17]. The Hologic Selenia Dimension system recently received the premarket approval from FDA [18].

From the technological point of view the current DBT scanners however have two important challenges: the relatively low spatial resolution and long scanning time. Although [Type text]

results from phantom imaging [19, 20] and clinical tests [21] have shown that DBT has a higher sensitivity for mass detection compared with mammography, DBT by itself is found to be inferior to diagnostic mammography in characterization of micro-calcification (MC) which is critical for diagnosis of cancer [22-24]. MC detectability in DBT can be affected by many factors including the detector type, reconstruction and acquisition parameters. Motion blurs due to source/detector motion and patient motion during acquisition is a dominate factor affecting the spatial resolution of DBT, thus the detection of small MCs [14].

The designs of the current scanners are based on a full-field digital mammography (FFDM) system. A full tomosynthesis scan may take from 4 seconds to more than 1 minute depending on the number of views acquired, the angular coverage, the total imaging dose and the x-ray tube power. In the current systems the scanning speed and the spatial resolution are interconnected [5, 25]. To generate the limited angle series of projection images, a mammography x-ray tube moves along an arc above the partially compressed breast either in step-and-shoot or continuous motion mode. In the former case the source needs to come to a complete stop at each position before x-ray exposure. The mechanical instability induced by acceleration and deceleration of the source limits the speed by which the tube can be moved from view to view [26]. In the latter case the x-ray tube moves continuously through the arc during the scan which introduces significant motion blur on the projection images. The higher the scanning speed the larger the distance the x-ray tube travels during the finite x-ray exposure time window and therefore the larger the focal spot motion blur [6, 27]. This effect is particularly severe for thick breast. The problem can in principle be mitigated by using a shorter x-ray pulse width with a higher power x-ray tube. However due to constraints on the anode heat load there is very limited room for increasing the x-ray power with the current rotating anode mammography x-ray tubes without sacrificing the x-ray focal spot size[28]. The Selenia Dimension uses a rotating x-ray tube operating in the energy range of 20-49kVp and tube current of 100-200 mA, and an x-ray focal spot size (FSS) of 0.45x0.75mm. To gain the x-ray photon flux and thus reduce the exposure time, a 0.7mm thin Al filter is employed to keep the tube output rate high. Because of the excellent low-dose DQE performance of the new selenium detector, the total dose could also be divided into more projections, which helps to reduce the x-ray dose, focal spot blur in each projection image and maintain adequate image sharpness. In the current configuration, 15 projections over 14 degrees scan angle is selected. The total dose is always divided equally among all projection views and the angular separation between projections is kept the same in a [Type text]

scan. With 100mAs total dose and 3.7s scanning time, the exposure time per view is 35ms, the motion blur along the tube rotation direction is 1.7mm.

The Selenia Dimension scanner has a very unique combo mode that allows both 2D and 3D breast images to be acquired quickly under single breast compression, thus results in co-registered 2D and 3D breast images. 2D conventional image has the advantage of better image sharpness while 3D tomosynthesis images can remove structure noises and improve mass lesion detection. The combo images of 2D and 3D together are expected to combine good calcification detection of 2D and good mass detection of 3D, to improve the detection sensitivity and to reduce the recall rate [13].

One way to increase the spatial resolution and reduce scanning time is to utilize a spatially distributed x-ray source to generate the projection views without mechanical motion. In stationary digital breast tomosynthesis (s-DBT) different projection images are acquired by electronically switching on/off the corresponding x-ray beams without mechanical motion of either the tube or the detector [29-31]. The motion blur associated with source motion can be completely eliminated, and the in-plane image resolution can approach that of FFDM. In this case the scanning speed and therefore the issue of patient motion is determined by the x-ray tube flux and the detector readout time. The concept of s-SDBT was demonstrated using a bench top system with mechanically assembled x-ray source array in a vacuum chamber and a Varian Paxscan detector in a geometry similar to that of the Siemens Mammomat scanner [32]. Projection images of a breast phantom were collected using the x-ray source array from 25 different viewing angles without motion.

Here we carried out a detailed evaluation of the feasibility of improving the spatial resolution of the Hologic's Selenia Dimensions scanner if the standard single-beam mammography x-ray tube is replaced with a distributed MBFEX array. The scanning times that can be achieved using the MBFEX source are evaluated and discussed for a given set of detector integration time and readout time. The evaluation was carried out for an imaging configuration and protocol closely matching that of the Hologic's Selenia Dimensions [13].

2. METHODS

The Selenia Dimension system can perform a 15 view scan with 14 degree angular coverage in 3.7s for an imaging dose of 100mAs. For the 30 degree coverage at 100mAs the scan

[Type text]

time becomes 7.4s. The focal spot motion blur in the 14 degrees and 30 degrees scans are respectively 1.7mm and 1mm.

The goal of this research is to investigate the feasibility of eliminating the x-ray tube motion blur while achieving 4 second scanning time for both 14 degrees and 30 degrees scans at 100mAs imaging dose in the same imaging geometry as the Selenia Dimension scanner. To answer this question the following studies were carried out: (1) electrostatic simulations and measurements to design an electron focusing optics to achieve the desired x-ray focal spot size; (2) thermal simulations to determine the maximum anode heat load; (3) electron field emission measurements of the carbon nanotube (CNT) cathodes to assess the maximum stable tube current that can be generated at the desired focal spot size; and (4) design of the x-ray source array.

2-1. Electron focusing optics

The Selenia Dimension system has an x-ray FSS of 0.45x0.75mm along tube motion direction and normal to tube motion direction respectively. Since x-ray tube rotates, the focal spot motion blur is 1.7mm for 15 view scan, 3.7s scanning time for an imaging dose of 100mAs. Therefore FSS with motion blur is 2.15x0.75mm. The calculated modulation transfer function (MTF) is 1.7x3.4 cycles/mm. The proposed UNC x-ray beam has FSS of 0.7x0.6mm, MTF is 4.2x3.8 cycles/mm, as shown in [Table1](#).

The structure to generate a single x-ray beam is shown in [Figure 1](#). The x-ray unit is composed of a carbon nanotube cathode, a gate electrode, two focusing units, and an anode. Carbon nanotube field emitters are deposited at a predefined small area on a metal substrate. 31 such assembled x-ray units were put into the constructed x-ray tube and are operated at high vacuum conditions.

The appropriate focusing optics design for the proposed UNC x-ray beam is based on computer simulations using a commercial software package (VectorField) to evaluate the modified Einzel lens with three active focusing electrode designs [33]. The first consideration was to determine the cathode current required per beam to get total dose of 100mAs. The second one was to determine a cathode emission area which can deliver the high current of 40mA reliably in a stable manner. The overall goal is to select the design that gives the required FSS and transmission rate with less design complications. From the previous study it has been confirmed that an elliptical cathode area of 2x8mm can reach 18mA cathode current in triode mode [32]. This was a clear indication that a larger emission area is needed to achieve 27mA
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anode current. A CNT cathode of 2.5x13mm rectangular shape was determined after careful considerations of emission area needed to get high current as well as sustainability of thermal loading by the gate mesh. The process of design optimization involved varying different parameters such as the distance between focusing electrodes. The opening apertures for the electrodes can also be varied. The geometric parameters were optimized to give desired FSS and transmission rate. Optimization of all the geometric parameters led to the final design is shown in Figure 1.

Table1: Specifications of the Hologic DBT-Selenia Dimensions and UNC s-DBT			
	Hologic without motion blur	Hologic with motion blur	UNC
FSS along tube motion (mm)	0.45	2.15*	0.7
MTF (cycles/mm)	4.3	1.7	3.6
FSS normal to tube motion (mm)	0.75	0.75	0.6
MTF (cycles/mm)	3.4	3.4	3.8
Motion mode	continuous rotation		stationary
* Since the gantry motion is continuous, its motion can affect spatial resolution. 1.7mm blur is based on 100mAs, 15° coverage, 15 beams, 3.7s scan time.			

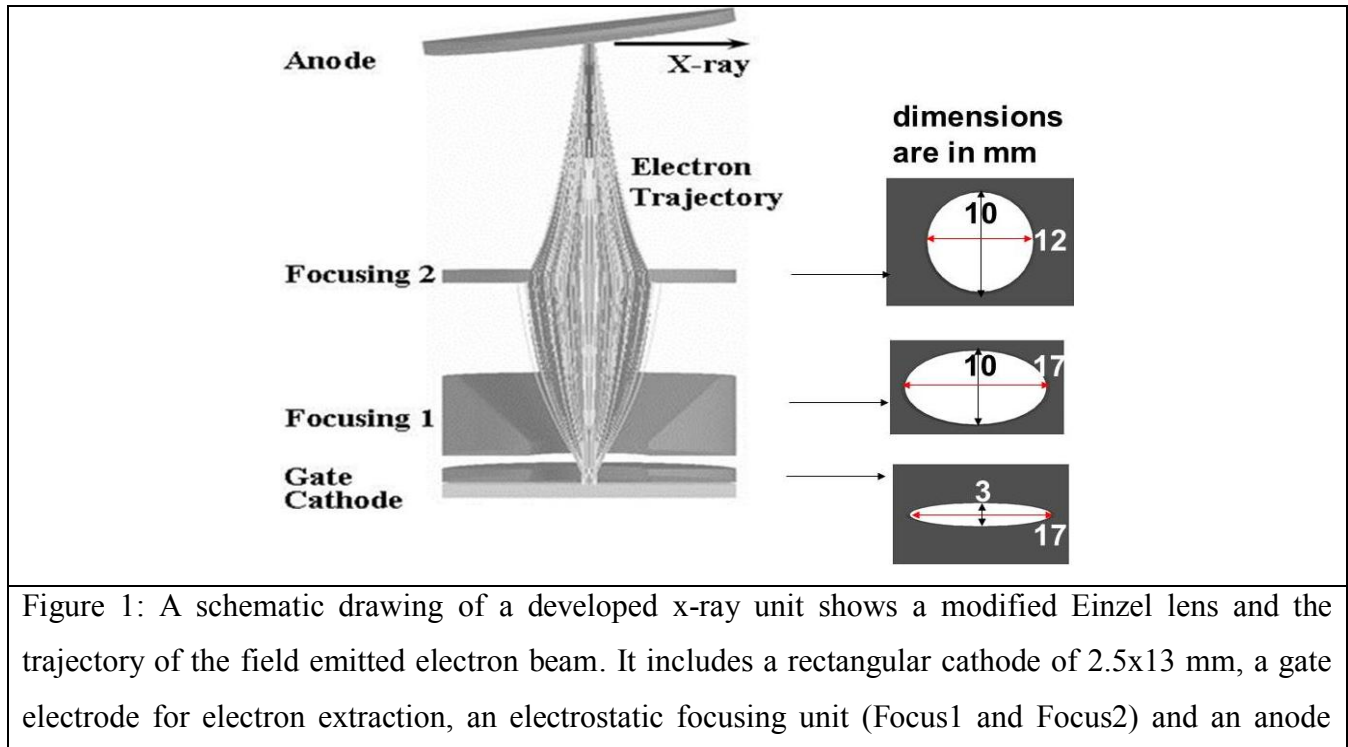


Figure 1: A schematic drawing of a developed x-ray unit shows a modified Einzel lens and the trajectory of the field emitted electron beam. It includes a rectangular cathode of 2.5x13 mm, a gate electrode for electron extraction, an electrostatic focusing unit (Focus1 and Focus2) and an anode

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made of Tungsten. The focusing electrodes are made of stainless steel and have elliptical apertures.

2.2. Anode heat load

The main source of thermal energy generation in a field emission x-ray tube, as well as a thermionic x-ray tube, results from bombardment of the high kinetic energy electrons on the metal anode. About 99% of the kinetic energy of the electron is converted to heat on the anode. A conventional rotating anode mammography x-ray tube operates at 100-200mA tube current at ~30kVp anode voltage, resulting in a peak power of 3-6kW. The rotating anode design is used to distribute the thermal load over a large area. In the case of the MBFEX source designed for the proposed s-DBT scanner, the anode is an extended metallic structure. The thermal energy is distributed over the full length and the anode block has sufficient heat capacity to absorb the thermal energy generated during the scan. The energy input of 100mAs x 40kV = 4kJ leads to a very small temperature rise compared with temperature rise in conventional x-ray tubes. The peak power of the MBFEX source is only ~1kW. This power has to be removed from the tube to allow a scan of every 10mins. However there is a concern whether the individual focus spot will be heated up too close to the melting temperature of the target material for the time duration the electron beam is on. For a 100mAs exposure with 15 projection views, each focus spot is exposed for 250ms and 125ms respectively for the 4s and 2s scanning protocols.

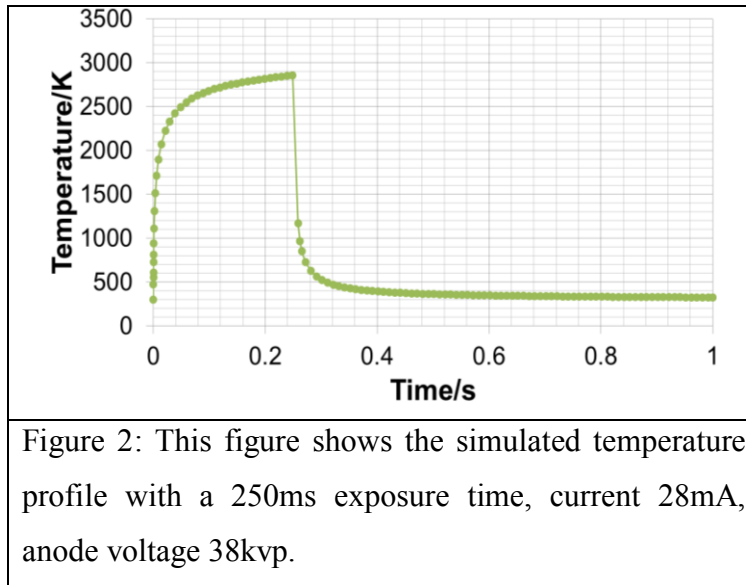
Finite element simulation is used to study the thermal energy dissipation with a tungsten anode. The temperature on anode can be determined by solving the heat equation:

where c_p is the heat capacity, k is the thermal conductivity, both c_p and k are temperature dependent, P_{in} is the input power and P_{rad} is the output power due to blackbody radiation.

The Transient Thermal Module on ANSYS 12.1.0 is used to simulate the heat load on the anode. The simulation is setup on a $60mm \times 30mm \times 5mm$ Tungsten anode with a focal spot size of $2.5mm \times 0.5mm$ (effective FSS $0.68mm \times 0.5mm$) and fully temperature dependent thermal parameters. Anode high voltage is 38kVp, current is 28mA. The power

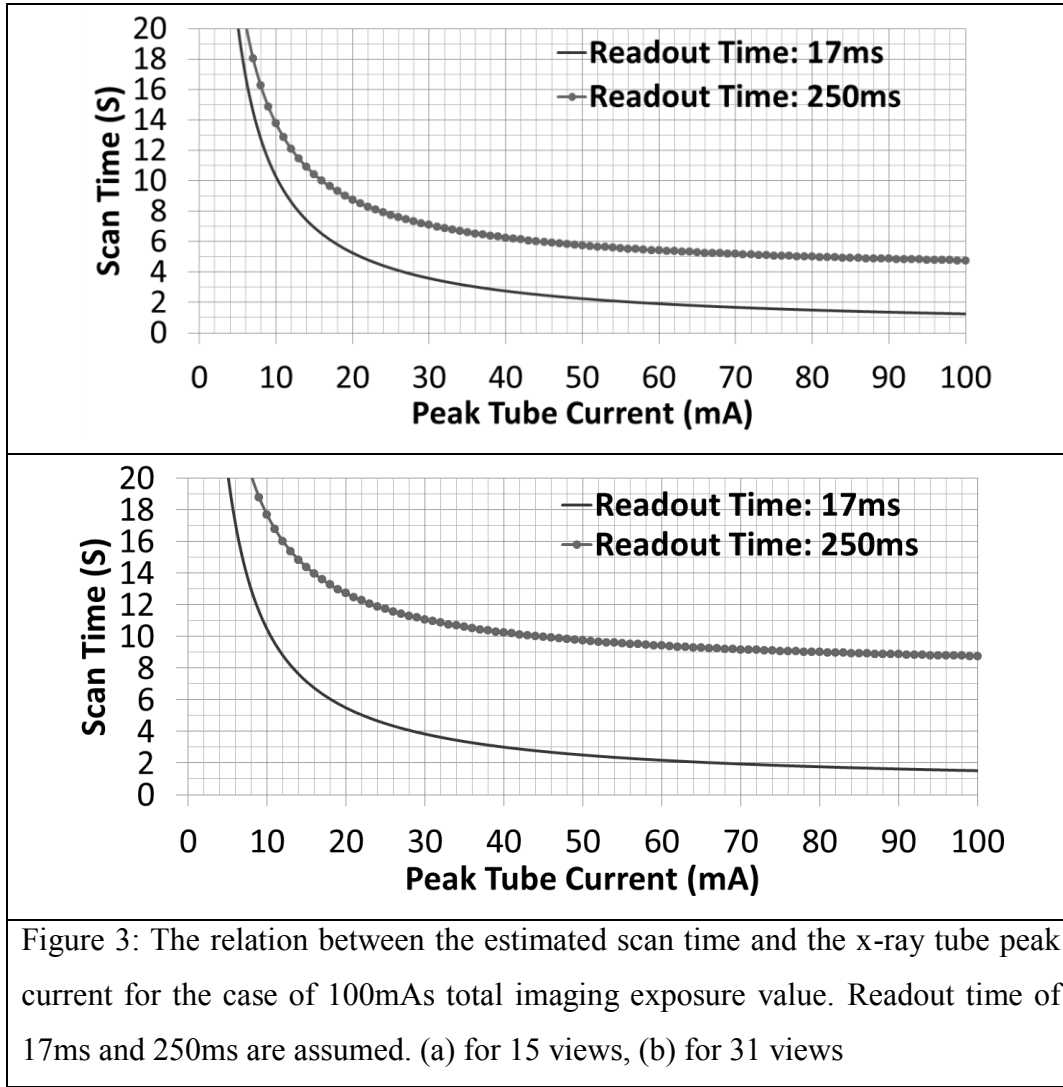
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density distribution at the focal spot is implemented with Gaussian distribution. Temperature profile is simulated with a 250ms exposure time. The simulation result is shown in Figure 2. Under the 250ms exposure, the anode can generally stand the heat generated. From the maximum temperature profile, we can see the anode temperature drops down quickly so that the anode is not affected much by the heat generated by accelerated electrons.



2.3. Scanning time and the x-ray tube current

For the s-DBT scanner, the scanning time (t_{scan}) is influenced by multiple factors including the number of views (N_{view}), detector readout time (t_{readout}), integration time ($t_{\text{integration}}$), total imaging exposure value (D_{total}), and x-ray tube current (I_{tube}). If the detector frame rate is not a limiting factor, then the scanning time can be calculated by: $t_{\text{scan}} = N_{\text{view}} \times (t_{\text{readout}} + D_{\text{total}} / (N_{\text{view}} \times I_{\text{tube}}))$. Assuming 100mAs is the total imaging exposure value distributed evenly over the 15 beams, the relation becomes: $t_{\text{scan}} = 15 \times (t_{\text{readout}} + 100 / (15 \times I_{\text{tube}}))$. The relation between the scanning time and tube current is plotted in Figure 3.



The objective scanning time in this study is 4s without motion blurring. This value compare favorably to the reported scanning time of the prototype scanner from commercial vendor [5, 6, 26]. With 17ms detector readout time, a tube current (per x-ray beam) of 27mA is required to achieve the 4sec scanning time in the 15 view protocol. In this case the exposure time for each beam is 250msec. To evaluate the feasibility of achieving a stable emission current of 27mA for the targeted focal spot size of 0.7x0.6 mm on the anode and with a pulse width of 250msec, CNT cathode with the correct dimension for the targeted focal spot size was fabricated by the electrophoretic deposition process developed in our lab. A multi-beam x-ray testing module was constructed for the testing entire x-ray beam structure with the cathode, the extraction gate electrode, focusing electrodes.

Our recent study showed the fabrication of high power-rating gate mesh is critical to achieve the high stable x-ray output power of the s-DBT scanner. The new explored gate mesh designs use Tungsten foil. The generated cathode current was measured using the Tektronix current measurement systems (TCPA 300, TCP312 and TDS3000C). The gate electrode with gate mesh attached was grounded and the cathode was connected to the negative output of power supply (Glassman ER3R100). The cable connected the Cytec switching system and power supply went through the current probe TCP312, the cathode current was therefore acquired when x-ray was on.

2.4. X-ray source design

The MBFEX source array is designed to provide a similar imaging configuration as the Hologic Selenia Dimension scanner, in terms of angular coverage, number of views, source-detector-distance, etc. as tabulated in [Table 2](#). The x-ray source array consists of 31 individual x-ray generating beams arranged in a linear array which is parallel to the detector plane, as shown in [Figure 4](#). The x-ray source array is housed inside a custom made stainless steel vacuum tube with a rectangular x-ray window. The x-ray window is covered by a 1mm thick Al foil which provides the vacuum seal and also serves as the x-ray filter. The x-ray anodes are tilted towards the center of the object (defined as the isocenter). [Figures 4](#) show the system geometry. The 31 x-ray beams in the x-ray tube span a distance of 37cm from end-to-end with 1 degree increment per projection. At a SOD of 65 cm, it provides 30° coverage (the linear spacing between the x-ray beams varies to provide the even angular spacing). The object center is considered as the ISO-center which is 5 cm above the detector.

The x-ray beams can be programmed individually in any time sequence to allow various imaging configurations. The three standard options are: 15 views using central 15 beams with 1° per projection and 14° coverage; 31 views with 1° per projection and 30° coverage and 15 views using 15 beams with 2° per projection and 30° coverage.

The x-ray beams are collimated to cover the whole flat panel detector with tolerance 2% of source-imaging-plane distance (SID) on three sides of detector except the chest wall side. On the chest wall side, beam edge is 5mm away from the chest wall. For an SID of 700mm the tolerance is 14mm. In the current pre-collimator design the exposed area exceeds the detector area by 5mm on each side. The chest-wall side is further collimated with an external collimator.

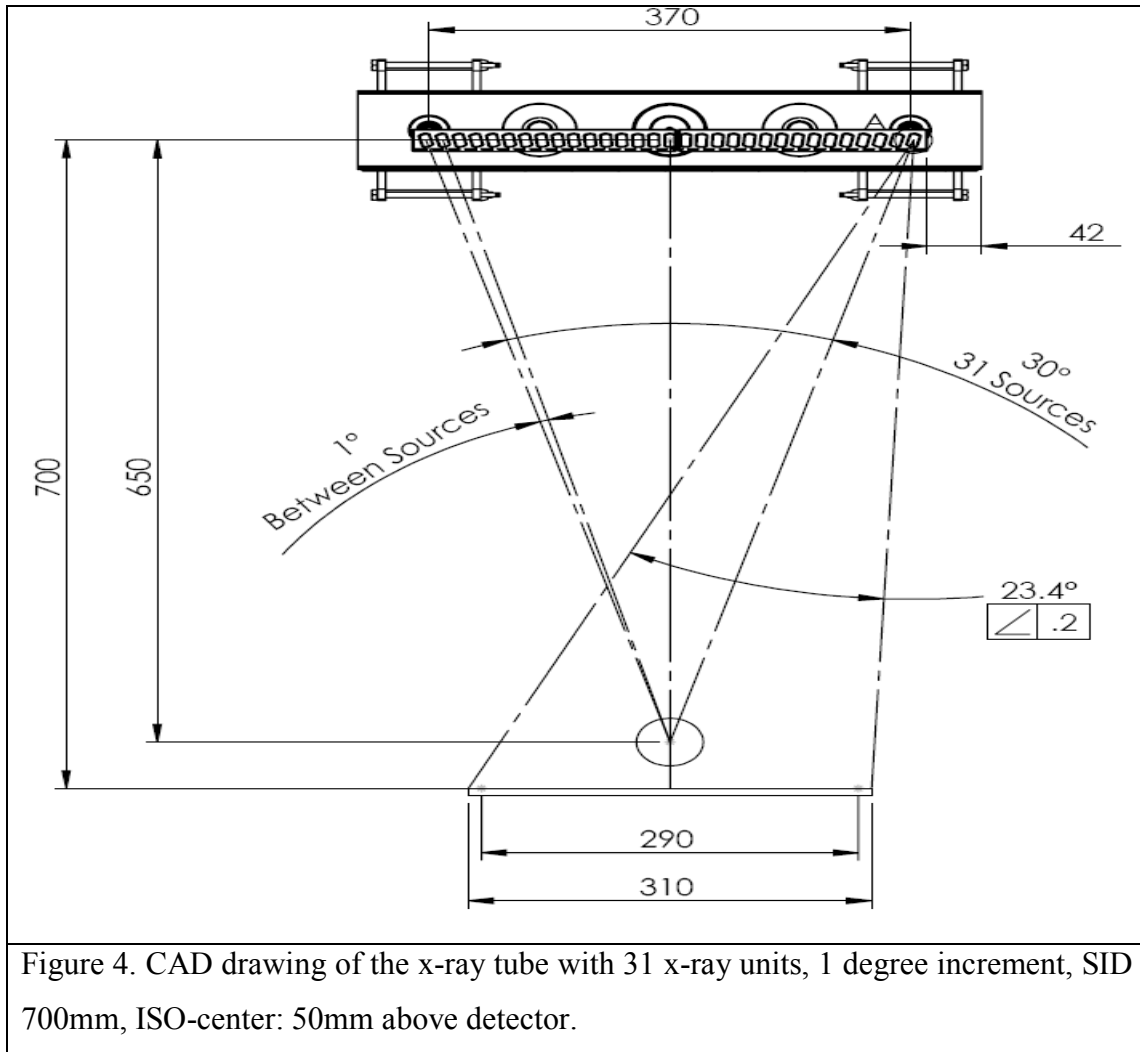
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To avoid heel effect and make sure the central part of the beam distribution is in the center of the detector, x-ray tube angle is 6 degree and anode angle is 16 degree.

In this parallel imaging geometry, there is a slight beam-to-beam variation in source-object-distance which can be calculated precisely from the location of the x-ray beam. If the distance from the central beam (#16) to object is SOD_{16} , the distance from other beams to object is $SOD_i = SOD / \cos((i-16) \cdot \theta)$, with beam number $i=1 \dots 31$, angular increment $\theta=1^\circ$, $SOD_{16}=650\text{mm}$. To make sure the entrance dose on the object from all beams is the same, the mAs from each beam is adjusted based on the SOD.

Table2: Specifications of the s-DBT scanner	
Number of views	15 - 31
Angular range	Up to 30 degrees
Detector FOV	24x29cm
Scanning time	4s
x-ray tube current	27 mA
Detector readout time	0.017s (2x2 binning)
x-ray anode voltage	28-38kV
X-ray anode/filtering	Tungsten/Al



2-5 Control Electronics

The x-ray unit array operates in a sequential mode where each x-ray beam is activated sequentially by switching on and off the connection to the CNT cathodes. This is accomplished through a computer controlled switching system (HXV/96, Cytec Inc), which has contact rating 200 watts, switch voltage 3500 volts, switch current 3.0 amps, maximum operating time 3.0 msec. HXV/96 switching system can contain 12 modules, each module has 8 reed relay [Type text]

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channels and a connector to cathode power supply. SHV cables are used to connect reed relay channels to CNT cathodes. Since we have 31 cathode beams, 31 reed relay channels from four modules are used. The transistor-transistor logic (TTL) trigger signal is provided by the selenium detector. Once the workstation receives the TTL signal from the detector, field-programmable gate array (FPGA) card (NI PCI-7830R) embedded in the workstation will synchronize both the detector and x-ray source to acquire images. To minimize the current fluctuation and decay, and to reduce source to source variation, an adjustable resistor box and an electrical compensation loop are also incorporated. Before running the s-DBT system, we can use the adjustable resistor box to compensate potential difference between gate and cathode to achieve constant tube current from each x-ray beam, then the electrical compensation loop will be used to dynamically adjust the exposure time to maintain a constant exposure mAs level. **Figure 5** shows that the detector trigger and current are switched simultaneously. In **Figure 6**, it shows that the current pulse is delayed 1ms consistently. The rising edge of the current pulse is sharp, while the falling edge of it has a tail of ~1ms. If exposure time from any beam is larger than 250ms, that means the current is low, the scan will be terminated and x-ray beams will be calibrated by adjusting the resistors.

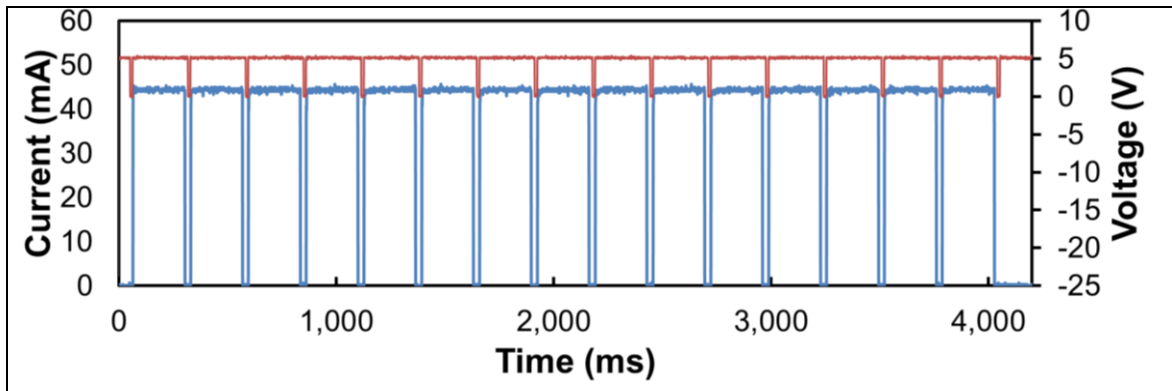


Figure 5. The upper pulse train is the TTL signal (5V) from detector controller, rising edge triggering. Pulse width is 250ms. Pulse width is adjustable. Current pulse train shows constant current level with varied pulse width which makes mAs slightly different among 15 beams. This is because the distance from 15 beams to the iso-center is varied.

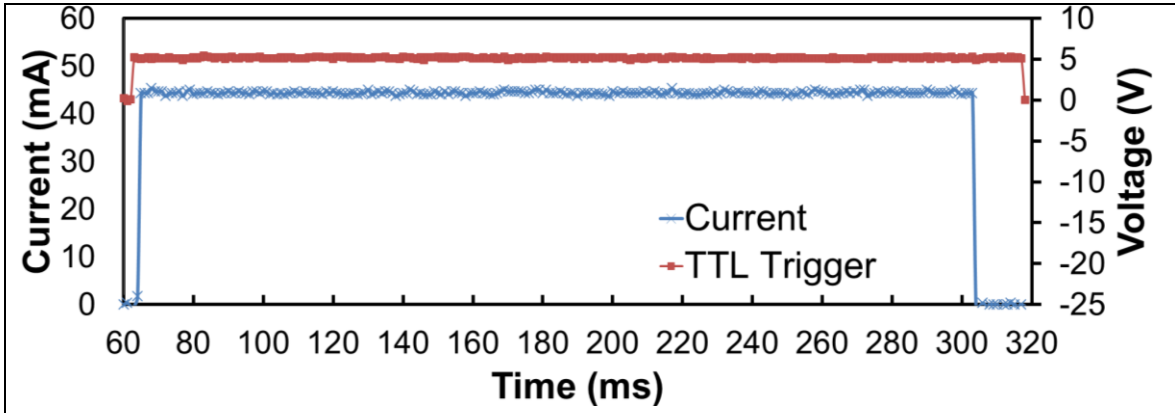


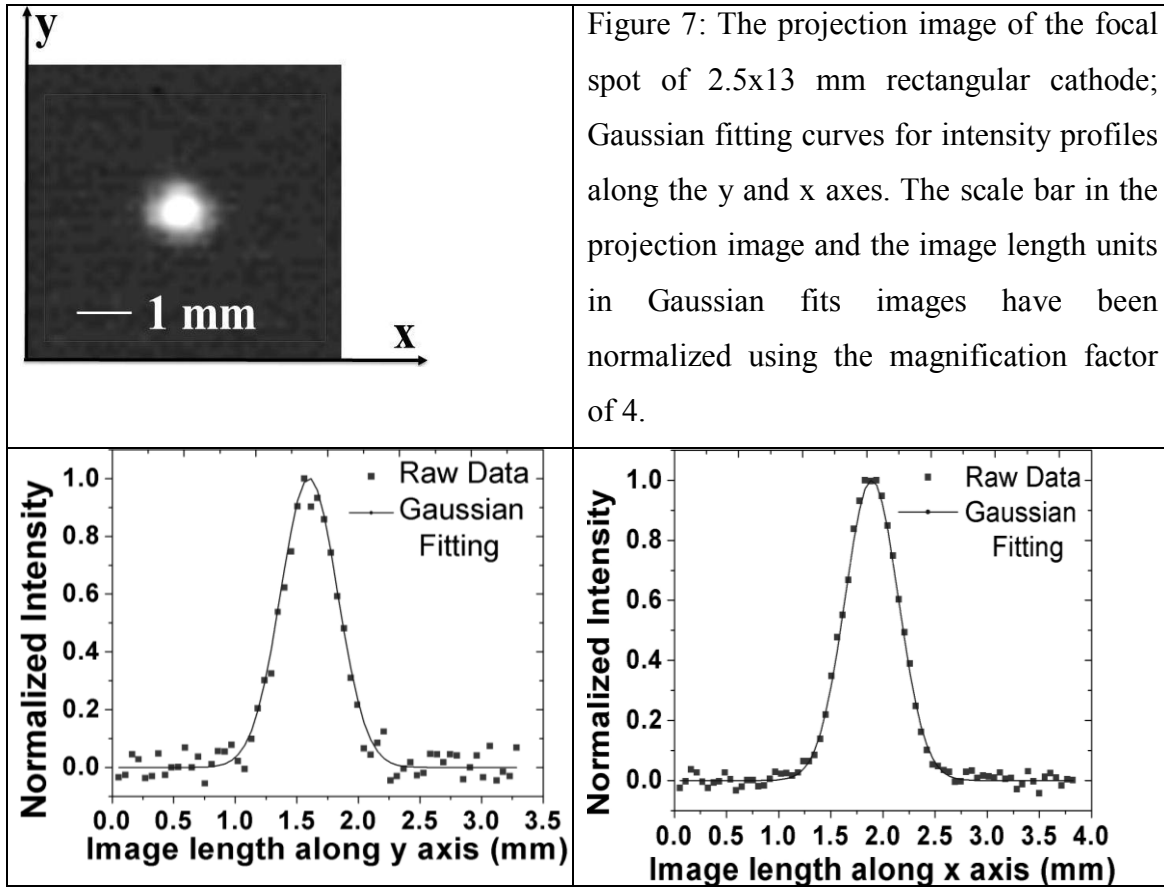
Figure 6. The current pulse is delayed 1ms consistently. The rising edge of the current pulse is sharp, while the falling edge of it has a tail of ~1ms.

3. Results

3-1 Focal Spot Size Measurement

Each x-ray generating unit comprises a CNT field emission cathode, a gate electrode to extract the electrons, and two modified Einzel-type electrostatic lenses to focus the field emitted electrons to a small area on a Tungsten anode. The anode, focusing electrodes and the cathode will be electrically connected to the following 4 power supplies respectively: Spellman SR70PN6 (70KV, 85mA), two Stanford Research PS350 (2.5KV, 10mA), and Glassman ER03R100 (3KV, 100mA).

The effective FSS of the x-ray beam has been measured using a gold-platinum pin-hole phantom. The diameter of the pinhole is $d=100\mu\text{m}$ with 500 μm length and 12° opening angle. The obtained FSS results are shown in Figure 7. The experimentally measured FSS of the $2.5\times 13\text{mm}$ rectangular cathodes is $0.7\times 0.6\text{mm}$, which is better than $2.15\times 0.75\text{mm}$, the FSS of Hologic DBT system with motion blurring. X axis is the x-ray tube linear array direction, y axis is the direction normal to the tube linear array. The FSS was measured regarding the IEC standard [34].

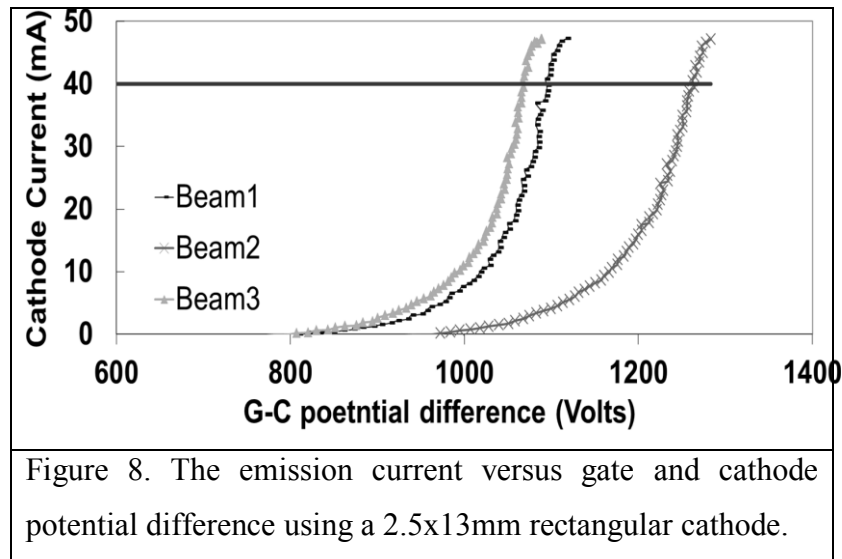


3-2 Experimental Demonstration of X-ray Source Stability

The cathode current measured recently in this study readily reached 40mA using 2.5x13 mm cathodes, as shown in Figure 8. The gate and cathode potential difference among three beams can be easily compensated by using our home made adjustable resistor box. Figure 9 shows the lifetime measurement results using 2.5x13 mm rectangular cathodes. The experiment was performed at peak cathode current 40mA, pulse width 250ms. The pulse waveform in the subfigure of Figure 9 shows the pulse shape of cathode current, the pulse width is 250ms. The current value is 3 x 40mA, it is because of the way we measured the current using the Textronics current probe. As we can see, the cathode shows very little degradation after 8000 minutes measurement time, the total x-ray on time is 424 minutes. Since the measurement data we acquired is too large to be plotted out using OriginLab software, we plotted three sections of the whole data: beginning, middle and last sections. The trend-line is used to connect these three sections. For an imaging protocol using 100mAs per tomosynthesis scan with 15 x-ray beams, the dose from each x-ray beam is ~ 6.7mAs, which translates to 10mAs per cathode using the [Type text]

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measured transmission rate of 67% (67% of the cathode current reached the anode). Assuming the tube is operating at the cathode current of 40mA (a stable emission at this level has been readily obtained at 1262 V applied cathode voltage), the required exposure time is 0.25s per cathode. The 424 mins electron beam-on time translates to an x-ray source array lifetime of at least 101760 scans. For a busy mammography clinic with ~ 60 patients per day, the results means that MBFEX source can be operated for at least 1696 working days at this output level. If we consider 250 working days per year, this MBFEX source can last for ~ 7 years. During measurement the gate voltage essentially remained the same, indicating the CNT cathode can be operated for far longer time than this. The lifetime measurement of the x-ray source is still under evaluation using higher peak tube current.



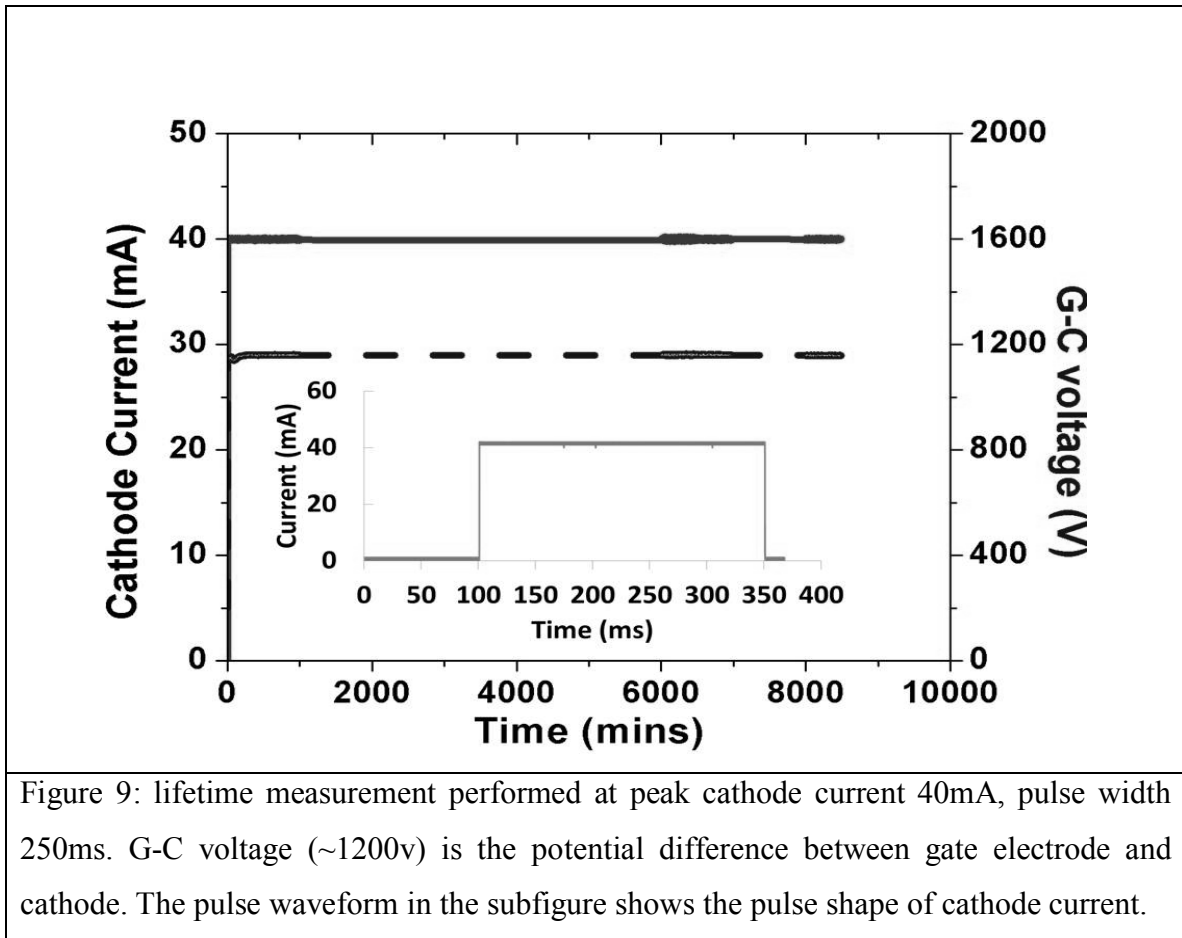


Figure 9: lifetime measurement performed at peak cathode current 40mA, pulse width 250ms. G-C voltage (~1200v) is the potential difference between gate electrode and cathode. The pulse waveform in the subfigure shows the pulse shape of cathode current.

3.3 Regulation of the x-ray flux from each beam

The CNT x-ray source array comprises 31 beams. Inevitably there is variation in the performance of the individual CNT cathodes which will result in variation in the output x-ray flux unless corrected. During the course of its lifetime the field emission properties of the CNTs will degrade which needs to be compensated to ensure the same x-ray output. Several steps are taken to minimize the beam-to-beam variation and to achieve the constant output during the lifetime: 1) an adjustable ballast resistor is connected to each CNT cathode to compensate the initial variation in the driving voltage; and 2) an electrical compensation circuit is built to automatically adjust the driving voltage needed to maintain the constant emission current (and therefore x-ray flux). Before running the CNT x-ray source array, calibration is first performed to determine the difference in the different CNT cathodes. The ballast resistances are set to compensate the potential difference between gate and cathode to achieve constant tube current from each x-ray beam. **Figure 10** shows the potential difference between gate and cathode among [Type text]

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31 beams, and resistor value for compensating this difference for each beam to reach targeted cathode current 43 mA. An electrical compensation method is used to dynamically adjust the exposure time to maintain a constant exposure mAs level.

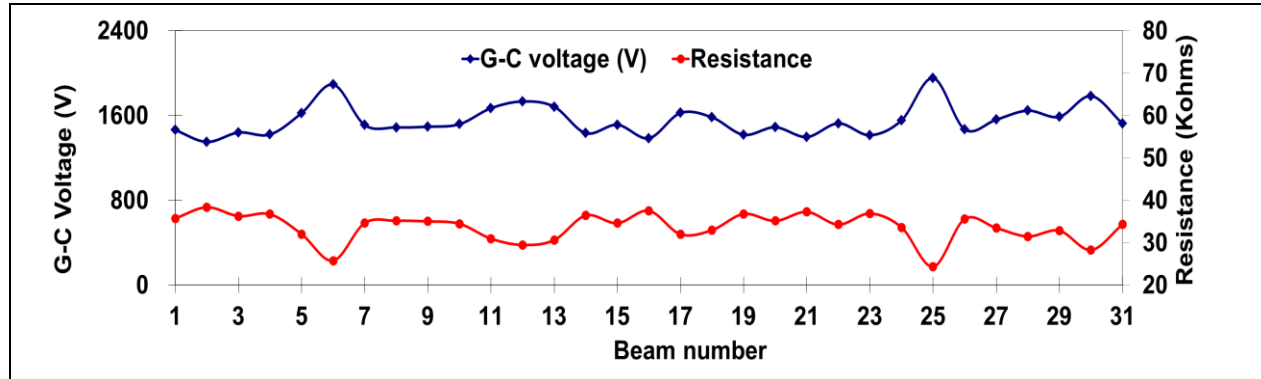


Figure 10. The upper curve shows the potential difference of gate and cathode, the lower curve shows the compensating resistance for each beam to reach targeted cathode current 43mA.

3.4 Initial breast phantom images

Figure 11 shows the raw projection image of a tissue-equivalent breast phantom (Model 013, CIRS, Inc.) using the central x-ray beam. The phantom body is shaped to represent a partially compressed breast, about 5cm in thickness. Embedded within the phantom are randomly positioned solid masses, and two MC clusters are in the center layer of the phantom. The image was obtained using: 28 kVp, 3.3 mAs per view. The masses and MCs in the phantom can be readily visualized in all 15 views.

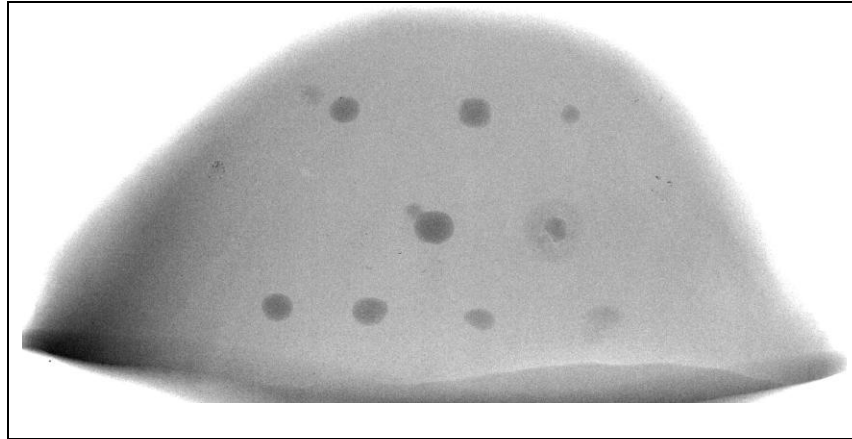


Figure 11: The raw projection image of the tissue-equivalent breast phantom using central x-ray beam.

4. Discussion and Conclusion

In this article, we have completely demonstrated our high temporal and spatial resolution s-DBT scanner. Using the experimentally measured current of 40mA (tube current of 27mA at 67% transmission), scanning time can be as fast as 4s in 2x2 binning mode for the total imaging dose of 100mAs. This value is comparable with the reported scanning times of Hologic DBT scanners. Further reducing the scanning time can be achieved by either a detector with fast frame rate and higher tube current. For example, with the current detector, if $I_{\text{tube}} = 40 \text{ mA}$ then t_{scan} becomes 5.7s at full resolution and 3.3s with 2x2 binning. Based on the emission stability from the current study, we believe a higher current can be achieved without enlarging the cathode size and therefore the FSS. Our recent study showed the fabrication of high power-rating gate mesh is critical to achieve the high x-ray output power of the s-DBT scanner. Recently we are exploring new gate mesh designs with different materials, such as Tungsten. The experiment was performed at peak cathode current 40mA, pulse width 250ms. The performance of the gate mesh is still under evaluation with higher peak tube current. The s-DBT design enables better spatial resolution compared to Hologic DBT scanner. The experimentally measured FSS of 2.5x13mm rectangular cathodes is 0.7x0.6 mm, which is favorable compared to that of Hologic DBT scanner with motion blurring. FSS is correlated to output power and heat load on anode. Given certain size of focal spot on anode, the heat load on anode increases accordingly with the increase of output exposure level. To maintain sufficient output exposure level for imaging, it is essential to keep heat load on anode within certain range (tungsten melting point is 3400°C). The initial projection image acquired using the central beam at 3.3mAs, 28KVP shows that s-DBT

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can provide high quality images.

These preliminary results demonstrate the feasibility of the proposed s-DBT scanner. The technology has the potential to increase the resolution and reduce the scanning time. The flexibility in configuration of the x-ray source array will also allow system designers to consider imaging geometries that are difficult to achieve with the conventional single-source rotating approach.

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Purpose: The current digital breast tomosynthesis (DBT) scanners are based on regular full-field digital mammography systems and require partial isocentric motion of an x-ray tube over certain angular range to record the projection views needed for reconstruction. This prolongs scanning time and degrades imaging quality due to motion blur. The purpose of this study is to evaluate the feasibility of improving spatial resolution and scanning speed of the current DBT device by replacing mammography tube with a stationary carbon nanotube (CNT) based x-ray source array.

Method and Materials: A spatially distributed multi-beam field emission x-ray (MBFEX) source array was designed for DBT. Electrostatic simulations were carried out to design the electron focusing optics to achieve the desired x-ray focal spot size. Finite element analysis was performed to determine the x-ray source anode heat load. A control electronic system was developed to scan and to regulate the imaging dose from each beam by compensating the driving voltages and by modulating the exposure time from each beam. The x-ray flux, lifetime, focal spot size, variation between different beams, etc were characterized. **Results:** A MBFEX source array with 31 individually controllable beams covering 30 degrees viewing angle was fabricated. It is operated up to 50KVp anode voltage and 30mA tube current per beam at an effective focal size comparable to that of the traditional mammography tube. Consistency in the beam-to-beam performance was achieved by optimizing the CNT cathodes and by electronic compensation. Tomosynthesis images of a breast phantom were obtained using the new source array. **Conclusions:** These preliminary results demonstrate the feasibility of CNT x-ray source array for stationary DBT to eliminate image blurring from tube motion. With a fast detector the device can provide a higher spatial resolution at a reduced imaging time compared to the current rotating gantry DBT systems.