Published in final edited form as:

IEEE Trans Nucl Sci. 2009 February ; 56(1): 36–45. doi:10.1109/TNS.2008.2008647.

Design and Development of a New Multi-Projection X-Ray System for Chest Imaging

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Abstract

Overlapping anatomical structures may confound the detection of abnormal pathology, including lung nodules, in conventional single-projection chest radiography. To minimize this fundamental limiting factor, a dedicated digital multi-projection system for chest imaging was recently developed at the Radiology Department of Duke University. We are reporting the design of the multi-projection imaging system and its initial performance in an ongoing clinical trial. The system is capable of acquiring multiple full-field projections of the same patient along both the horizontal and vertical axes at variable speeds and acquisition frame rates. These images acquired in rapid succession from slightly different angles about the posterior-anterior (PA) orientation can be correlated to minimize the influence of overlying anatomy. The developed system has been tested for repeatability and motion blur artifacts to investigate its robustness for clinical trials. Excellent geometrical consistency was found in the tube motion, with positional errors for clinical settings within 1%. The effect of tube-motion on the image quality measured in terms of impact on the Modulation Transfer Function (MTF) was found to be minimal. The system was deemed clinic-ready and a clinical trial was subsequently launched. The flexibility of image acquisition built into the system provides a unique opportunity to easily modify it for different clinical applications, including tomosynthesis, correlation imaging (CI), and stereoscopic imaging.

Index Terms

Chest radiography; correlation imaging; lung cancer; stereoscopy; tomosynthesis

I. Introduction

Lung cancer is the second leading cause of mortality in the US after cardiovascular disease. An estimated 160 390 lung cancer deaths occurred in 2007 alone [1]. One of the reasons for this high mortality is the fact that many patients present with advanced stage disease, when potentially curative surgery is no longer possible. In order to decrease lung cancer mortality, it needs to be diagnosed earlier.

Early stage lung cancer may manifest as a solitary pulmonary nodule, prior to development of evident mediastinal or distant metastases. It was therefore suggested that screening patients at risk for lung cancer with chest radiographs might facilitate detection of early stage disease and reduce mortality. However, all trials that used chest radiographs to screen for lung cancer showed no change in overall lung cancer mortality and, as a consequence, chest radiographic screening for lung cancer was not recommended [2], [3].

Perhaps one of the reasons that chest radiographic screening failed to reduce lung cancer mortality is the fact that the sensitivity and specificity of chest radiography for the detection of lung nodules is relatively low and that up to 90% of potentially cancerous pulmonary nodules are not detected (“missed”) on chest radiographs [4]–[6].

One of the major deficiencies of chest radiography is its limitation in rendering suspected pathology that a radiologist may be looking for. Cancerous nodules may be hidden in a complex cloud of overlapping projections of the anatomical features of the thorax such as mediastinum, pulmonary vessels, lung structure, the heart, and diaphragm resulting in false negatives [7], [8]. The detection of lung nodules could potentially be improved by an imaging technique that can minimize the undesirable influence of overlying anatomy. An ideal approach to eliminate overlapping anatomy is Computed Tomography (CT). However current utilization of CT as a screening tool for identifying lung nodules on a broad basis is controversial because of economical (cost and technology availability), patient care (e.g., overdiagnosis), and patient dose considerations [9]. The radiation dose to a patient undergoing a CT scan, even a low dose CT scan, is much higher than that in chest radiography, and thus, from a public health standpoint, the higher associated health risks might not be justifiable for a wide-scale CT lung cancer screening program. Thus if lung cancer screening is to be realized, there is a need for cost-effective and dose-effective imaging alternative.

Towards that end, multi-projection imaging has been proposed as a technique to improve early stage cancer detection over standard projection technique without increasing the current dose levels [10]–[12]. In multi-projection chest imaging, a plurality of digital radiographic images of the same patient are acquired in quick succession from slightly different angles within a limited angular range. The total dose, which is equivalent to that of typical chest radiography, is equally distributed across all the projections. By exploiting the differences in geometrical perspectives that different angular projections offer, the multi-projection imaging scheme is able to reduce the confounding effects of anatomical overlap that otherwise limit the efficiency of chest radiography. Improved detection of potential lesions is obtained by combining information from the multiplicity of angular projections.

Practically, this combination can take different forms including scrolling of the projection images manually or in cine mode, stereoscopic display of projection images [13], computer-aided analysis of the images [14], or tomosynthesis [15].

A multi-projection x-ray system for chest imaging was recently designed and developed at the Radiology Department of Duke University. The purpose of this paper is to present physical design considerations in the development of this system and to report the translational steps that advanced the system from the developmental stage to its final clinical implementation.

First, different components in the development of the system will be described, followed by tests conducted to investigate its precision in image acquisition. These tests were conducted since robust image acquisition is pivotal to the clinical performance of the system. Finally, different clinical applications of the multi-projection x-ray system such as tomosynthesis, correlation imaging (CI), and stereoscopic imaging will be noted.

II. Methods

A. The Imaging System Hardware

A standard x-ray system with upright gantry was modified to enable the x-ray tube to translate along the horizontal and vertical axes in an iso-centric motion with the center of the x-ray beam located at a distance of 2.5 cm in front of the detector. In either direction, the x-ray tube was rotated about the focal-spot of the x-ray beam to complete the iso-centric configuration. The detector is stationary, amorphous silicon indirect (CsI) flat panel sensor (Paxscan, 4030CB series; Varian Medical Systems, Palo Alto, CA) designed to perform at high frame rates with extended dynamic range. It has a pixel size of $194\ \mu\text{m}$ and a matrix size of 2048×1536 .

Motion along the horizontal and vertical axes is made possible through two linear actuators driven with servo-motors. While a similar system has been built by Dobbins *et al.* [16], the unique feature of this system is that the x-ray tube is capable of moving in-synch along both axes, and therefore can traverse an arbitrary trajectory on a plane parallel to the detector (Figs. 1 & 2). Another differentiating feature is the easier mechanical setup of the system; it uses a single servo-motor in conjunction with a simple lever and railing arrangement to provide motion along the vertical axis. The railing, shown in Figs. 2 and 3, is angulated such that it provides a tilt of 1° for every inch of tube translation resulting in a maximum angular span of $\pm 7.5^\circ$ about the posterior-anterior (PA) orientation along the vertical axis. This approach is less costly and easier to construct than the alternative methods. Motion in the horizontal plane and the articulation required to provide fulcrum of the x-ray beam upon the detector are provided by two additional servo-motors enabling a maximum angular span of $\pm 15^\circ$ along the horizontal axis.

B. Synchronization and Electronics

A single controller unit synchronizes motion along the two axes and the rotation of the x-ray tube by executing user-commands from a software platform (Motion Planner, Cross Automation, Belmont, NC). Acquisition of images is orchestrated between the image

acquisition software (ViVA, Varian Medical Systems, Palo Alto, CA) and the user to expose the detector by the x-ray generator. The motion is also regulated by hardware interrupts generated by the end-of-travel hardwired switches fitted on to the actuators (Fig. 2). The switches also facilitate position-error checks. This is done as follows: the switches, which are placed at locations that mark known coordinates along the trajectory of the tube-motion, trip when the tube passes them during its motion. These trip signals (hardware interrupt signals) are in turn compared with the real-time coordinates of the tube that are inquired by the software. Any misalignment between the expected and the actual coordinates of the tube is compensated by the motion controller unit. This error check is automatically performed at end of each acquisition, thereby compensating for any possible misalignment in the position of the tube and correctly resetting it for subsequent acquisition.

A custom software-based imaging protocol synchronizes the acquisition task between the acquisition software, x-ray generator and the positional coordinates of the x-ray tube along the two axes. A schematic of the interface developed to achieve the synchronization is shown in Fig. 4. Fig. 5 depicts the timing diagram of the synchronization scheme. The labels A through I indicate the chronological order of the constituting system tasks leading to final image acquisition. Sequence A indicates the desired frame rate set by the user on an external frequency generator. Upon activation (Sequence B & C), the x-ray generator and the image acquisition software trigger the detector and set the image acquisition frame rate of the detector. The read-out and reset/dead time of the detector is approximately 60 ms and is independent of the acquisition frame-rate. The detector generates a pulse in which the falling edges signify an active low-state when the detector is ready to acquire images (Sequence D). This pulse serves as an input interrupt for the tube motion-controller (initiated by the user in Sequence E) and synchronizes the image acquisition software with the tube motion (Sequence F). The first falling edge of the detector pulse initiates the tube translation (Sequence G). At the second falling edge, the motion controller sends an active high pulse to the x-ray generator (Sequence H). At this instant, the x-ray generator, which is triggered active by the logical AND of the falling edge of the detector pulse and the active high pulse from the motion-controller, generates x-rays thus exposing the detector (Sequence I). The detector, which is in the active read-in state at this instant, starts acquiring the image.

This acquisition protocol ensures that no image frames are lost after the patient is exposed to radiation, avoiding unnecessary patient exposure. The images are acquired at the frame-rate set by the user in Sequence A. The translating speed of the tube is adjusted to synchronize the image acquisitions with the desired positional coordinates of the tube. Once the desired number of images in the acquisition sequence has been acquired and the tube has translated across the desired angular span, the motion controller sets the trigger off, which in turn stops the x-ray exposure and hence, image acquisition.

The system is capable of acquiring multiple images from different angles in rapid succession in both step-and-shoot and continuous modes. The readout-rate of the detector is user adjustable with a maximum rate of 7.5 frames per second at full-detector resolution. The x-ray tube can be programmed to translate at variable tube speeds in order to optimize total acquisition time for superior image quality and maximum patient comfort.

C. Geometrical Precision

Having developed the system, its geometrical robustness was investigated to determine its suitability for clinical implementation. The geometrical robustness was assessed in terms of the precision in position and possible motion blur due to tube motion. Toward that end, multiple projection images of a slightly angulated tungsten edge-device were acquired with the tube in motion (continuous mode) at the clinically relevant kVp setting of 120 for chest radiography. In addition, 19 mm of A1 filtration was added to emulate spectrum attenuation through patients. An edge device was placed at a distance of 30.5 cm from the detector to represent the maximum geometrical magnification expected from patients [17]. A small focal spot size (0.6 mm, normal) was used to minimize the effect due to focal spot blurring. Exposure times were 10 and 5 ms. The tube was translated at 2.5 and 5 cm/sec and images were acquired at 1 and 2 frames/sec, respectively, resulting in two combinations of tube speed and frame rate settings. Total acquisition times were 10 and 5 seconds, respectively, times anticipated in actual clinical trials. These images acquired at different settings were used to investigate the geometrical robustness of the system, as described next.

For assessing geometrical precision of the system, five series of 15 image acquisitions were obtained at each of the two combinations of tube speed and frame rate settings. To evaluate the positioning/velocity errors encountered during translation of the x-ray tube, the position coordinates of each of the 15 images reported by the motion planner were compared across the different sets of images obtained at each of the two tube-speed/frame rate setting combinations. Projections of the edge device acquired at the PA orientation with and without tube motion were compared to evaluate any error in the reported positional coordinates.

To investigate possible motion blur artifact along horizontal axis, the expected PSF of motion blur for a point at a distance of z from the detector was calculated analytically. Based on the geometry of tube trajectory, the PSF was determined to be $zvt/(SID - z)$, where v is the tube speed, t is the exposure time and SID is the source-to-image distance. Furthermore, using images of the edge device, the Modulation Transfer Function (MTF) was computed experimentally with the tube in motion and compared with that obtained with stationary tube. A previously published methodology [17] was used to compute the MTF from the images of the edge device.

D. Clinical Trial

Having developed the multi-projection imaging system and determined the robustness of its image acquisition capability, a clinical trial with Institutional Review Board approval, including written informed consent, has been initiated to image patients with confirmed lung nodules and normal volunteers. So far, 80 out of a total target of 250 subjects have been imaged. To identify patients with lung nodules, the CT databases at our institution are searched for recently detected or diagnosed malignancies (either primary or metastatic).

For each patient, three projections are acquired along the horizontal axis. These are acquired at $+3^\circ$, 0° (PA orientation) and -3° . The tube settings are such that each of the three images is acquired at 1/3rd the dose level of a typical chest radiography examination. Specifically, the tube voltage is fixed at 120 kVp, and the tube current/exposure time product is varied

between 1.25–6.4 mAs (in order to maintain consistent image noise across varying patient chest thicknesses in the 20–36 cm range). The images are acquired in a continuous acquisition mode with the tube speed of 2.5 cm/sec and at a read-out time rate of 1 frame per second resulting in total acquisition time of 10 seconds—within a patient's single breath-hold. The source-to-image distance (SID) is 200 cm, corresponding to the optimum magnification based on an earlier study [18]. A small focal-spot size (0.6 mm, normal) is used to minimize the effect due to focal spot blurring.

Before scanning the patients, the system is calibrated for a precise geometry of the x-ray tube translation. The calibration takes into account the SID, the speed of the translating tube, and the acquisition frame rate. This ensures image acquisition at the desired angular orientations. Specifically, the tube location is first reset to the central PA position. An automatic error check, as described earlier, is performed at this point to ensure precise alignment of the tube with the desired trajectory. Using the geometry shown in Fig. 1 and the known SID, the tube is then displaced by 13.8 cm which corresponds to a displacement of $+3.8^\circ$ relative to the PA position. This position is at a slight offset from $+3^\circ$ -location of the first acquisition, and was provided so that the tube motion stabilizes before the onset of x-ray exposures. Next, at user's prompt, the tube motion is initiated per sequence G of Fig. 5, and synchronized image acquisition is initiated as noted earlier. Following the three acquisitions, the tube is repositioned to the PA position thus completing the acquisition cycle.

The system is operated under the supervision of a dedicated clinical coordinator and technologist. Once the subject is ready for acquisition and is positioned in front of the detector, the coordinator prompts the technician to initiate the custom-built automatic routine rendered on an easy-to-use graphical interface. The routine automatically initiates the scanning procedure, scans and exposes the subject, and also acquires and saves the images on a dedicated computer.

III. Results and Discussion

A. Image System Performance

The system was found to have an excellent geometrical precision in terms of the reported positional coordinates of the translating x-ray tube at four different overall combinations of tube speed and frame rate settings. The positional errors at tube speeds of 2.5 and 5 cm/sec were found to be within 1% and hence inconsequential. However, images captured with the stationary tube revealed a slight misalignment from those captured at the same position when the tube was in motion. This may be due to a delay of approximately 100 ms between the time when the motion-controller reports the coordinates and the actual position at which the x-ray generator initiates the exposure. The motion-controller reports the positional coordinates at the instant when it receives an active ready-to-acquire falling pulse from the detector. This falling pulse simultaneously causes a voltage to be applied across the cathode and the anode of the x-ray tube. We speculate that there may be approximately a 100 ms delay in the time required by the x-ray tube to generate steady-state x-ray beam after application of this voltage. Consequently, after the positional coordinates are reported, the tube translates to a new position, resulting in a misalignment. However, at typical tube

velocities in the 2.5–5 cm/sec range, the resultant misalignment will be 0.02–0.05 mm range, and hence negligible for actual clinical acquisitions.

Fig. 6 shows MTFs acquired with exposure times of 5 and 10 ms at tube speeds of 2.5 and 5 cm/sec. The overlying MTFs indicate the image quality was not affected by increase in exposure time at either tube speeds, and also remained unchanged when the tube was moved in the negative or the positive direction with respect to PA orientation. To be sure, the expected drop in the MTF due to motion was calculated analytically. It was found for a point at a distance of 30 cm from the detector, the expected drop was ~1% at the Nyquist frequency (2.57 lp/mm), thus confirming the overlying nature of the experimentally-determined MTFs. Since the method implicitly assumes a linear shift invariance property of the system, the measured MTFs were independent of the slight error in positional coordinates reported by the motion-controller. Moreover, at 2.5 and 5 cm/sec, effect on spatial MTF due to tube motion was found to be minimal. It may be noted that this analysis was based only on the horizontal axis—the axis employed for our clinical trial. Although not directly applicable to the vertical axis, the MTF evaluation representative of the performance and stability of the system and indicates minimal effect of tube motion on overall image quality.

To be sure of the nominal effect of tube motion in our clinical trial, PA image of a complex thoracic phantom (Kyoto Kagaku. Co., Ltd, Kyoto, Japan) was acquired with the tube moving at 2.5 cm/sec corresponding to the tube speed during clinical trials and compared it with the PA image acquired with static tube. These are shown in Figs. 7(a) and (b), respectively. The insets show magnified view of high-frequency components around a rib edge. In comparing the two images, it is clear that the image acquired with the tube in motion faithfully represents high frequency components, and thus preserves anatomical details. There is practically no difference between the two images, thus visually confirming the minimal impact of tube motion on the clinical image quality.

B. Clinical Implementation

With the prototype multi-projection system set up for clinical implementation, multi-projection images of 80 subjects have been successfully acquired as a part of our ongoing clinical trial. The clinical trial is scheduled to run for another year, until approximately 250 expected subjects are enrolled.

Images of all the 80 subjects imaged so far were judged to have excellent diagnostic image quality by experienced radiologists. No artifact due to the system itself was observed in the acquired images. Fig. 8 shows sample clinical images of three subjects acquired at $+3^\circ$, 0° , and -3° about the PA orientation. While the top row shows images from a healthy volunteer, the bottom two rows show images of patients with the location of the suspected lesions identified and marked with arrows. Most noteworthy is the relative change in the location of the lung nodules across the three views. This information of the spatial displacement of the identified lesion between different views combined with the knowledge of geometry of acquisition may be used by a computer algorithm to identify lesions and reduce false positives [10]. Each pair may also be evaluated stereoscopically [19]. The analysis and report of the clinical utility of the system awaits completion of the human subject study.

C. Clinical Applications

The multi-projection imaging scheme, by virtue of its intrinsic limited-angle tomography mechanism, reduces the main limitation of standard projection imaging of overlapping anatomical structures (i.e., anatomical noise) that can partially or completely hide pathology of interest. Using this method, multiple images of the same patient are acquired within a short time interval, and the correlation of information in multiple projections is used to arrive at the final diagnosis. This information may be harnessed by an image-processing application that can take advantage of the multi-projection setup. The unique features of our multi-projection imaging system enabling the acquisition of images at variable tube speeds, frame rates, and orientations provides the potential for a variety of multi-projection applications, including tomosynthesis, CI, and stereoscopic imaging, briefly discussed below. It may be noted that not all of these applications have been implemented in our laboratory, however, the installed hardware can potentially support all of the following applications.

Tomosynthesis—Tomosynthesis is a realization of multi-projection imaging scheme in which the projection images are further processed to compute images parallel to the detector plane across varying depths within an organ of interest. Interrogating one image plane at a time minimizes the influence of anatomical structures of neighboring planes, resulting in improved diagnosis [11], [12].

The flexibility in image acquisition built into our system allows for implementing tomosynthesis acquisition geometry. Specifically, the system allows for iso-centric acquisitions along an angular arc of $\pm 15^\circ$ and $\pm 7.5^\circ$ along the horizontal and vertical axes, respectively, with number of images in the 3–80 range. Such a scanning geometry also enables the proposed acquisition for optimal tomosynthesis image quality [20]. As a proof of concept, a pilot tomosynthesis acquisition of the Kyoto thoracic phantom was conducted. Specifically, 30 images were acquired spanning $\pm 15^\circ$ along horizontal axis and used to reconstruct a volume of slices using a shift-and-add algorithm. Fig. 9 shows two example slices. In comparing these images with the standard PA image shown in Fig. 7, it is clear that the slices removed some of the overlying anatomy present in the PA image. The slices rendered anatomical features in greater details, indicating the robustness of the system for future tomosynthesis-based applications.

Furthermore, by combining horizontal motion with the vertical motion, our system lends itself to implementation of arbitrary image trajectories of the tube motion. As a result, rather than the standard iso-plane trajectories of existing tomosynthesis data acquisition scheme, these novel image trajectories could take advantage of the dual-plane configuration of the system to implement arbitrary image trajectories to potentially derive superior diagnostic image quality in tomosynthesis imaging.

Correlation Imaging—Correlation Imaging (CI) is another realization of multi-projection imaging scheme in which the acquisition protocol is similar to a tomosynthesis technique except that the images are directly analyzed instead of being reconstructed, thereby avoiding the confounding effects of the reconstruction technique inherent to tomosynthesis [21], [22].

Each angular projection is acquired at a lower dose level so that the total patient dose is within the bounds of an optimized acquisition. These images are then processed by a computer algorithm that utilizes spatial correlation information between different angular projections to identify and positively reinforce the lesion signals between different projections [14], [23], [24]. CI has applications in chest radiography as well as breast imaging [25].

We also envision the use of CI as an adjunct to tomosynthesis imaging, with no added patient dose. The image processing technique developed for CI may be applied to the projection images acquired as a part of tomosynthesis imaging, providing the radiologist with another diagnostic opinion.

A derivative of CI is bi-plane correlation imaging (BCI) where only two projections are needed. In our implementation of BCI (per the clinical trial noted earlier), three images are acquired along the horizontal orientation. Images are paired and analyzed for assessing the correlation of signals in individual views. Preliminary patient data show acceptable diagnostic quality of the acquired images (Fig. 8). Exhaustive studies are underway to assess whether the system could be a beneficial diagnostic tool for improved detection of lung nodules [24], [26], [27].

Stereoscopic Imaging—Images captured by the multi-projection imaging system may also be paired and presented to a radiologist to be viewed stereoscopically. Stereoscopic view is accomplished by presenting the left and the right eye-views to the observer simultaneously. These two views are then processed and combined by the brain upon which a radiologist interprets the object with a perception of depth, thus allowing him/her to sort through the overlapping tissues for potentially improved detection.

There are a number of display techniques for stereoscopy. In our implementation of the technique, stereoscopy is achieved by a combination of two medical-grade liquid crystal display (LCD) set and a special pair of polarizing glasses that the viewer wears. The two eye-views are rendered at full-resolution on the two LCDs. These two displays are configured to be 110° apart and are bisected by a semi-transparent mirror (Fig. 10). The mirror serves dual-purposes of reflecting the image of the display above it and transmitting the image of the display below it. The mirror also polarizes these images in such a manner that the reflected and the transmitted images reaching the observer are resolved only by the right and the left lens of the polarizing glasses, respectively, thus achieving stereoscopic rendition.

A recent study of breast stereo-imaging showed improved clinical performance over the standard mammographic technique [13]. A similar implementation of stereoscopy for chest radiography may provide a three-dimensional view of the thorax enabling accurate and intuitive depth perception of the anatomical structures within the thorax, thus alleviating the effect of overlapping structures in the detection of lung nodules.

IV. Summary

We have described the design, development, testing, and clinical implementation of a new x-ray chest imaging system for implementing multi-projection imaging. Preliminary clinical trials indicate that the system, used as a diagnostic tool, could potentially be utilized to improve detection of lung nodules. Because of the flexibility the system provided in image acquisitions, it lends itself to a variety of applications such as tomosynthesis, correlation imaging, and stereoscopic imaging, each offering distinct advantages for improved nodule detection.

Acknowledgments

This work was supported in part by the Breast Cancer Research Program, Pre-Doctoral Traineeship under Grant W81XWH-06-1-0449 and in part by the National Institute of Health under Grant R01-CA109074.

The authors would like to thank Eric Porter, Eric Smith, David Fullerton, and Duane Copeland at the Duke Hospital for their hardware-related contributions for the development of the prototype system. They would also like to thank Anne Jarvis, Robert Saunders, Ben Pollard, and Xiang Li for their help in coordinating the clinical trial. Thanks to Jim Dobbins and Christina Li at DAILabs for generously providing their reconstruction routine for tomosynthesis evaluation. Thanks to Richard Youngblood for valuable editorial comments.

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Fig. 1. Schematic of a new multi-projection x-ray imaging system capable of multiple oblique-angled acquisitions about the PA orientation for chest imaging. (a) top-view showing the system's trajectory along horizontal axis, and (b) side-view showing motion along the vertical axis.

Fig. 2. Snapshot of the multi-projection x-ray imaging system installed at the Radiology Dept. of Duke University as applicable for chest radiography. Also shown are motion-regulating switches (indicated by arrows) to achieve iso-centric motion of the tube along vertical axis.

Fig. 3.

Pictures of the multi-projections system depicting its trajectory along vertical (a–c) and horizontal axis (d–f). The tube motion is depicted via pictures of the tube at three different locations (PA and two extremities) along the two axes. Also shown is the angulated railing arrangement and the servo-motor coupling (pointed by the arrows in c & d) that enable the iso-centric motion of the tube along vertical and horizontal axes, respectively.

Fig. 4. Schematic of the interface developed to achieve synchronization of image acquisitions between the detector, x-ray generator and the positional coordinates of the moving x-ray tube. The corresponding timing diagram is shown in Fig. 5.

Fig. 5.

Timing diagram for continuous acquisition. The system is capable of both step-and-shoot and continuous operations.

Fig. 6.

Comparison of MTFs computed from images of an edge device acquired at PA orientation with the stationary tube and with the tube moving in the negative direction with a velocity 2.5 cm/sec (a), in the positive direction at 2.5 cm/sec (b), in the negative direction at 2.5 and 5 cm/sec (c), and at 2.5 cm/sec with exposure times of 5 and 10 ms (d).

Fig. 7.

Example PA images of a complex thoracic phantom (Kyoto Kagaku. Co., Ltd, Kyoto, Japan) acquired with (a) and without (b) tube motion. The tube speed was fixed at 2.5 cm/sec corresponding to the speed used in our clinical trial. Anatomical details rendered in the image acquired with static tube are preserved in that acquired with the tube in motion, indicating minimal impact of tube motion on clinical image quality.

Fig. 8.

Sample clinical images of three volunteers acquired at $+3^\circ$ (a), 0° (b), and -3° (c) about the PA orientation using a recently developed multi-projection chest imaging system. While the top row shows images of a healthy volunteer, the bottom two rows show subjects with lung nodules. The locations of the nodules were identified by a radiologist and are denoted by arrows. Most noteworthy is the relative displacement of lesion locations across different views. This information is used by multi-projection imaging technique to improve the accuracy of nodule detection.

Fig. 9.

Tomosynthesis slices of the Kyoto thoracic phantom as a proof of concept for potential tomosynthesis application of the new multi-projection system. The precision in image acquisition of the system enabled tomosynthesis acquisition resulting in slices that rendered anatomical features in greater details.

Fig. 10. Stereoscopic display unit at Duke Laboratory that enables 3D rendition of medical images, such as chest x-rays.