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Effects of Spatial Resolution and Spectral Purity on Transvenous Coronary Angiography Images

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Abstract

Measurements have been made on the National Synchrotron Light Source (NSLS) Coronary Angiography X17B2 beamline under ideal and real imaging conditions to investigate the optimal imaging conditions for spatial resolution and spectral purity. The spatial resolution tests were performed using two multielement Si(Li) detectors (600 element, 0.5mm pixel-pixel spacing; 1200 element, 0.25mm pixel-pixel spacing [1]). Images were taken of phantoms containing iodine contrast agent over a wide range of incident beam absorption conditions. Patient images were also obtained using the same viewing projection with both detectors.

Harmonics present in the imaging beam can be reduced by operating the superconducting wiggler source at reduced field strength. At regions of high absorption in the patient, the harmonics present can contribute to the detected signal [2]. Iodine

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phantom images were obtained at a wiggler field strength of 3 Tesla ($E_c=13.3\text{keV}$) and 4 Tesla ($E_c=17.8\text{keV}$) for comparison. As before, patient images were obtained using the same projection at both wiggler fields. Results of the detector resolution and wiggler field measurements will be presented for the phantoms as well as the patient scans.

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Introduction

The transvenous coronary angiography project at the National Synchrotron Light Source has obtained research quality images of human coronary arteries. In an attempt to optimize the imaging parameters, measurements have been made on the coronary angiography, X17B2, beamline under ideal and real imaging conditions to investigate spatial resolution and spectral purity. These measurements were comparisons between two horizontal resolution detectors and two operating fields of the superconducting wiggler source. The purpose of the detector change is to compare the resolution of a 600 element - 0.5mm pitch dual array detector with a 1200 element - 0.25mm pitch dual array detector. The field change was from 3.0T ($E=13.3\text{keV}$) and 4.0T ($E=17.8\text{keV}$) which in turn altered the harmonic content in the imaging beam. The presence of the harmonics in the beam add a background to the images acquired above and below the iodine absorption edge and degrade the logarithmic subtraction technique used to obtain an iodine image of the coronary anatomy. The phantom measurements were of an iodine containing phantom used to characterize the ability of the detector to resolve various dimension tubes. The phantom also contained lead strips which were used to determine the detector's spatial response to a step transmission function. Finally, patient images were taken under as nearly as possible the same conditions to compare each detector and wiggler field. These patient images were taken on subsequent days, with the detector swap occurring on one day and the wiggler field swap occurring the next.

A major logistical aspect of performing these system tests using patients was the time required to make the necessary changes in detectors or wiggler field. The patient's physician required that the time spent in performing the swaps not exceed 45 minutes. This time limit was to ensure the patient's safety and comfort. For this reason, the detector and field swaps were planned carefully and practiced several times to ensure they could be done quickly and without mistakes. In practice, both the detector and field changes were accomplished in under 30 minutes.

Spatial Resolution

As part of the upgrade program presently underway, a new detector and data acquisition system are being constructed [3]. The present detector has a pixel-to-pixel spacing of 0.25mm. The properties of the detector have been investigated and are reported elsewhere[1]. Similar measurements characterizing the older 600 element detector have also been performed. However, several changes in the data acquisition system[1] and in the beamline optics [4] have made a direct comparison between these two detectors extremely difficult. To ensure that the proper choice had been made in the construction of the new detector system, a decision was made to dedicate part of an angiography imaging run to resolving the uncertainties regarding the merits of each detector.

The present 0.25mm pitch detector has a total of 1200 elements with 600 elements in each row for the high and low energy beams covering a horizontal width of 150mm. Similarly, the 0.5mm pitch detector has a total of 600 elements in two rows of 300 elements each also covering a horizontal width of 150mm. The 1200 element detector allows higher spatial resolution, however, the increased resolution results in a loss of

signal in each pixel by a factor of two as compared to the 600 element detector. This loss of signal will increase the noise in the iodine image by approximately a factor of $\sqrt{2}$ if the noise is just due to photon counting statistics. It would appear then that the loss of signal to noise might be worth the potential gain in spatial resolution.

The Detector Swap

A hinged mount was prepared for the 600 element detector to allow it to be flipped in or out of the beam just upstream of the existing fixed 1200 element detector. A kinematic adjustment on the mount allowed alignment of the detector to the imaging beams. The mechanical swapping of each detector could be accomplished in a few minutes.

The signal cables from each detector use the same format cable and connectors. Each cable contained 12 twisted pair signal lines carrying signals from 12 detector elements. The 600 element detector had 50 such cables and the 1200 element detector had 100 cables. The signal cables were changed at the data acquisition system digitizer inputs. When the 1200 element detector was in use all of its 100 cables occupied all of the digitizer inputs. Since the 600 element detector signals required only half of the digitizer inputs, half of the cables of the 1200 element detector were always left in place. Swapping detector signals then amounted to swapping 50 ribbon cables in the data acquisition rack. Finally, to complete the change, the appropriate lookup table which maps the digitized signals into an image was loaded depending on the detector being used. For convenience, when the 600 element detector was used each pixel was mapped to two image pixel locations to simulate the same image size as the 1200 element detector. This allowed all of the existing software used to acquire, display and analyze

the 1200 element images on the data acquisition PC and subsequently on the workstation to remain unchanged.

Spatial Resolution Results

An iodine tube phantom was imaged repetitively at each of the 7 detector gain settings in two orthogonal orientations in order to assess the vertical and horizontal response to features in the phantom. Each successive gain setting increases the amplification of the signals by a factor of two. This phantom was imaged again after the swap of the detector. The iodine tube phantom was constructed from Plexiglas with various diameter holes drilled through it. The tube diameters were, 12.7mm, 3.18mm, 1.60mm, 1.02mm, 0.79mm, 0.51mm, and 0.34mm. Attached to the phantom were two lead tape crosses. These crosses were composed of two 10mm x 30mm x 0.25mm thick lead tapes orthogonally placed. This gave several edges in the vertical and horizontal direction which could be used to determine the step or edge response of the detector. It was from these edges that the modulation transfer function (MTF) could be determined.

The use of the edge response to determine the MTF[5] was used earlier[1] to characterize the 1200 element detector. Basically the procedure used here was to extract lines of raw low energy beam image data across the Pb tape edges in the vertical and horizontal direction. A numerical derivative of this edge response was taken to give the line spread function from which the MTF was determined by taking the Fourier transform. As found in the earlier measurements of the 1200 element detector the MTF of either detector does not depend on the attenuator in the incident beam and the detector electronics gain within the error of the measurements. Figure 1 shows the horizontal MTF for both the 1200

(solid line) and 600 (dashed line) element detectors up to their respective Nyquist frequencies. Also the vertical MTF is shown for both as dotted lines. As expected the vertical MTF's are very nearly the same since the image line data acquisition rate and scanning speed were the same for both detector systems (4.17msec/line and 12cm/sec scan speed, respectively). The superiority of the 1200 element detector can be seen by comparing the spatial response at the cutoff frequency of the 600 element detector at 1 line-pair/mm. The MTF of the 1200 element detector is 0.8 as compared to 0.6 for the 600 element detector. This value remains valid at imaging beam absorption values which approximate and exceed typical human imaging absorption conditions (measured over water equivalent thickness values from ~9 to 21cm).

A patient was imaged with both detectors using the same projection (Left Anterior Oblique (LAO) 30° projection) and with the same iodine contrast injection parameters (35ml Hexabrix at 15ml/sec). Both sets of images were taken at a 4.0Tesla wiggler field and the ring current was similar (242mA for the 600 element detector and 231mA for the 1200 element detector). Two image frames were taken in both cases on the eighth and ninth chair scan with iodine injection occurring on the second chair scan. The mean time between chair scans is 2.2seconds. Corresponding regions of the images acquired are shown in figure 2 for the image taken on the eighth scan. Figure 2a shows the overlap region acquired with the 600 element detector and figure 2b show the same region taken with the 1200 element detector. The difference in horizontal resolution is not readily apparent, however, upon closer inspection, smaller features are resolvable in the 1200 element detector image.

Field Swap

Patient and phantom images have been acquired at several operating fields of the superconducting wiggler (3.0T, 4.0T, 4.4T and 4.7T). Increasing the wiggler field will increase the harmonic content in the imaging beam. Since the attenuation of the higher energy components in the beam is significantly lower than that of the fundamental energy, the transmitted beam will contain a larger relative harmonic content and will degrade the difference image[2].

Since the installation of the bent Laue monochromator has dramatically increased the imaging flux which can exceed our administrative limit (7rads/frame skin entry dose), the wiggler field has been dropped to limit the maximum flux rather than using filters to lower the intensity. This allows flux reduction without the unnecessary enhancement of harmonics due to filtration. For this reason the most recent patient imaging sessions have been performed at 4.0T which is reduced from the normal operational field of 4.7T.

The detector, which operates in charge collection mode, cannot discriminate between the charge released from photons at the desired 33keV energy and multiples of that energy. The monochromator used to prepare the 33keV beam has little efficiency at 66keV but at 99keV the efficiency is relatively high. The imaging beam flux at 99keV relative to the 33keV fundamental is 0.45% at 4T and 0.10% at 3T. After traversing 20cm of tissue this relative harmonic content is altered to 9.6% at 4T and 2.2% at 3T. The relative response of the detector per photon at 99keV to that at 33keV is 25%. With this factor included the measured signal through 20cm of tissue will contain 2.4% 99keV harmonic contamination at 4T and 0.6% at 3T. This additional signal will degrade the ability to

determine absolute iodine value in the image. This in turn degrades the visual appearance of the image. At an iodine areal mass density of 10mg/cm^2 , which is larger than the expected values in the coronary arteries ($\sim 2\text{mg/cm}^2$), the calculated error in determining this value due to presence of harmonics is -3.4% at 4T and -0.8% at 3T.

Field Swap Results

To assess the effect of harmonic contamination, absolute iodine areal densities were measured at regions in the iodine tube phantom where the areal mass density should have been 10mg/cm^2 and compared with the error estimated above. The results of the measurements are shown in figure 3. The agreement between the calculation and the measured densities is poor, however, the trend of the measured data with increasing water equivalent thickness follows the calculation. Overall the error in the iodine determination is larger for the 4T field as expected.

Patient images were also taken at each field with the same projection for comparison. These images do not show a marked difference and the comparison was complicated by an unexpected increase in patient's heart rate. This resulted in the image being taken at the second field of 4T after the peak concentration of the contrast agent had already passed through the coronary arteries.

Conclusion

The measurements have shown that the spatial resolution of the 1200 element detector does in fact exceed that of the 600 element detector under real and ideal imaging

conditions. Based on this result the upgraded detector will maintain the presently used 0.25mm pixel pitch.

The results of the field change from 4.0T to 3.0T are not so conclusive. From the iodine phantom measurements there does appear to be some advantage in imaging at the lower field, however, the appearance of the phantom and patient images do not seem to be greatly affected by the change.

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Figure 1. Modulation Transfer Function (MTF) comparing the 0.25mm horizontal pitch, 1200 element detector (solid line) and the 0.5mm horizontal pitch, 600 element detector (dashed line). For reference the vertical MTF is shown for both at the dotted lines.

Figure 2. Patient images taken with the 600 element (a) and 1200 element (b) detectors. These are sections of the original images showing the distal right coronary artery along the bottom with part of the left ventricle and aorta in the upper right.

Figure 3. Measured and calculated iodine error plots for 3.0T and 4.0T wiggler fields at an iodine areal density of 10mg/cm^2 . The solid line is the calculated percent iodine error versus water equivalent thickness for 3.0T wiggler field. The open boxes are the measured iodine errors at this areal density and field. The dashed line shows the calculated iodine error at 4.0T and the crosses are the measured iodine errors.