HIGH RESOLUTION X-RAY COMPUTED TOMOGRAPHY FOR COMPOSITES AND ELECTRONICS INSPECTION

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BACKGROUND

A useful computed tomography (CT) or digital radiography (DR) system must simultaneously resolve both the smallest meaningful features and least relevant differences in density in the objects it is designed to inspect. The advent of structural composites and high-density electronic assemblies requires digital x-ray systems that are tailored to their properties. The critical flaw size in graphite composites is thought to be 25 to 50 μm. Similarly, prototype military electronics are now being assembled on circuit boards with conductive patterns only 100 μm in width.

Extension of digital x-ray radiography and computed tomography to this growing class of high-performance parts requires a detector that combines improved spatial resolution with other performance advantages including soft energy x-ray sensitivity, low interelement cross talk, reduced noise, and small physical size.

SYSTEM DESIGN

The system described here is a laboratory test bed for high-resolution digital x-ray imaging built at the GE Corporate Research and Development Center in Schenectady, New York (Figure 1). It is a third generation, fan beam scanner that has been demonstrated in both DR and CT imaging modes.

The system is configured on adjustable mountings so that different source-to-detector lengths and image magnifications can be easily achieved. However, because the x-ray emitting spot and the detector elements are both 25 to 50 μm wide, most imaging has been done with the target midway between the tube and the detector (Figure 2). The unsharpness [1] of the image as a result of focal spot size is

$$\text{unsharp} = \text{spot} * (D1/D2)$$
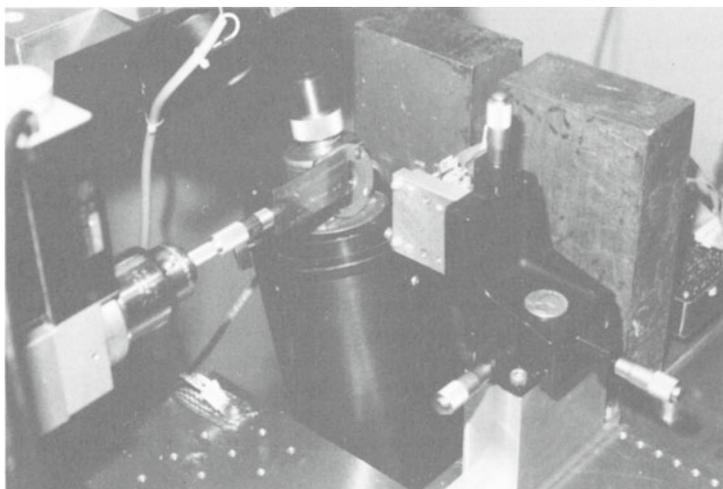
and, in our case,
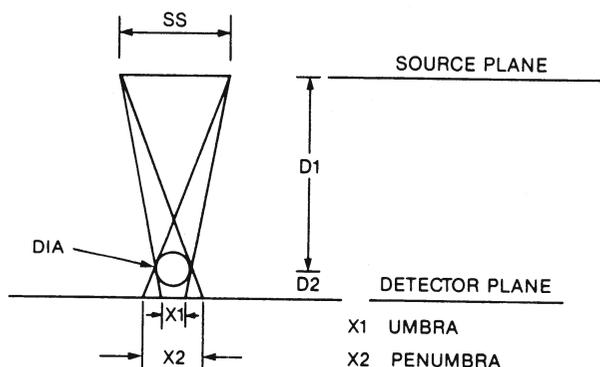
$$D1 = D2$$

Figure 1.   X-ray imaging system.

Figure 2.   Source and detector contributions to system resolution.

This magnification 2X geometry makes the unsharpness caused by the tube equivalent to the resolution limit of the detector and permits us to minimize the overall tube-to-detector spacing. This geometry makes optimum use of the modest x-ray flux available. The x-ray source is a Ridge 160 microfocus tube operated at 15 to 20 W continuous output.

The part manipulator is a 3-axis (x,y,θ) device with 0.1-μm accuracy over a 50-mm range. The axes are driven with a Aerotek Unidex 11 numerical controller interfaced to the system host by GPIB.

X-ray data are collected with an analog-to-digital converter and buffer memory constructed as CAMAC modules. The CAMAC transfers measurement results to a dedicated MicroVAX host. CT reconstructions are executed on a STAR array processor equipped with back-projection hardware.

DETECTOR

The heart of both this system and the development program that has surrounded it is a family of line array detectors tailored to the inspection of composites and electronics (Figure 3). These prototypes employ a new scintillator material to convert the x-ray flux to visible light and conventional integrated photo diode arrays to produce an electrical signal proportional to the instantaneous x-ray flux.

There is a substantial body of published experience with high-resolution x-ray imaging based on scintillation detectors. Much of the effort invested in these devices is targeted at improving the tradeoff between efficiently collecting the available x rays and decreasing the cross talk between adjacent detector elements. A thick scintillator is inherently better at stopping all x-ray flux. Unfortunately, the thicker materials also provide more opportunity for the scintillation light generated in one location to stray into an adjacent area. This decreases the contrast, which is required for high-resolution imaging.

In 1985, Seguin et al. [2] described a CT scanner for laboratory animals that included a detector consisting of a monolithic slab of scintillator material bonded to a linear array of many photodiodes. The authors attribute reflections in the scintillator block to long-range cross talk among the channels, which was removed mathematically. Other workers [3,4] have used both scintillating fiber optics and honeycomb structures to constrain the scintillation light in high-resolution x-ray imagers.

In the GE design, interchannel cross talk is reduced by physically dividing the scintillator material into individual bars that are optically isolated by reflectors or opaque layers. The approach is similar to the construction of scintillation detectors used for medical CT scanners, but on a substantially smaller scale. Our scintillator blocks are approximately 25 μm wide.

The effect of a physically divided scintillator is illustrated in Figure 4. The detector used for this radiograph was a CCD video image chip with a scintillator plate cut into long parallel bars bonded directly to the photosensitive area. A pair of identical crossed wires was projected onto the scintillator with a small medical x-ray source. The wire mounted parallel to the scintillator bars casts a crisp, high-contrast shadow on the image
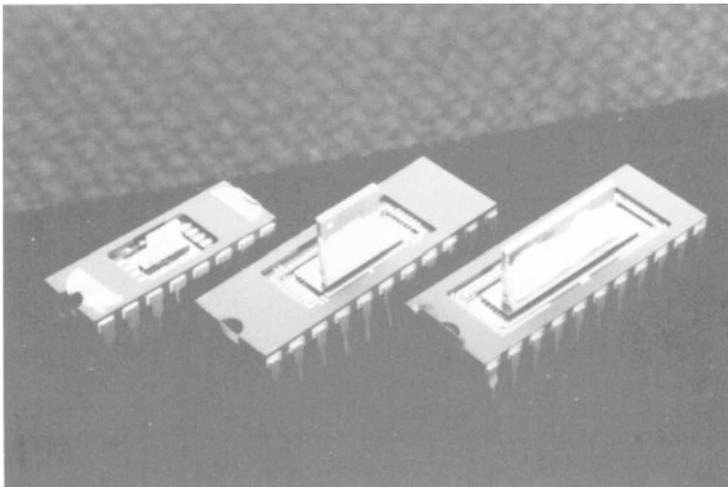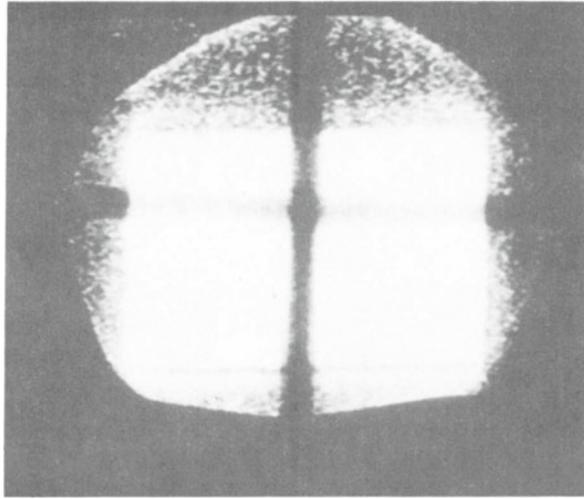

Figure 3.   High-resolution detectors.

Figure 4.   Zurich crosshair.

sensor because light from other parts of the scintillator material cannot efficiently leak through the opaque glue that separates the long bars. The image of the wire perpendicular to the scintillator bars does not benefit from the cross talk reducing divisions in the detector material. Light from other regions easily spreads along the bars of clear material and reduces the contrast of the wire shadow.

Our divided scintillator detector design is shown schematically in Figure 5. It consists of a linear array of scintillator blocks arranged in columns. The blocks are coated in reflective material on all sides except the bottom. Each block is one pixel or element in the linear array detector. The blocks are held in an evenly spaced line by adhesive attachment to one or more backing plates.
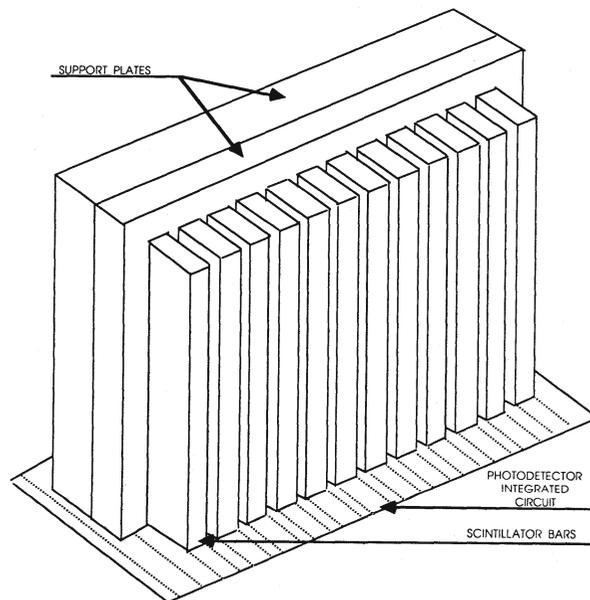

Figure 5.   X-ray detector array.

The bottom surface of the assembled scintillator blocks and their support is optically flat and polished. It is bonded directly to the silicon surface of an integrated circuit array of photodetectors which forms the basis of the detector.

In an imaging application, this device is positioned behind a slit collimator made of a dense material like tungsten or lead (Figure 6). The collimator permits x rays to hit only a section of the scintillator bars a fraction of an inch above the plane of the integrated circuit. The x rays that hit in this section generate visible light, which is reflected down to the integrated circuit, converted into an electrical signal, and ultimately transferred to a computer to form the final image. The collimator protects the integrated circuit from damage by the direct x-ray beam. The distance between the scintillating region and the integrated circuit protects the silicon device from scattered x-ray damage.

The integrated circuit detector array consists of a line of individual photosensitive areas and the circuitry necessary to read the cells individually. Most of our detectors are built on Reticon photodiode arrays.

At the current state of the art, we can produce arrays of scintillator material in which the bars are 2000 μm high, 25 μm wide, and 100 μm deep. These bars are mounted in an optically isolating matrix on 50-μm centers. Arrays up to 1.2 cm long have been evaluated so far.

This design enjoys fundamental advantages in sensitivity, contrast, and compactness. The high-density GE ceramic scintillator used in our prototypes stops more x rays in less volume than either gas or low-density scintillators. Furthermore, the material used here produces more light at a color better matched to silicon integrated circuit sensors than other media. These factors combine to produce a small device with high signal efficiency.

The reflecting media surrounding each scintillator bar efficiently channels its scintillation light into the photodiodes attached to it. There is only one glue interface between the scintillators and the photodetectors. This design reduces the cross talk and coupling between elements to lower levels than commonly seen in other scintillator designs.
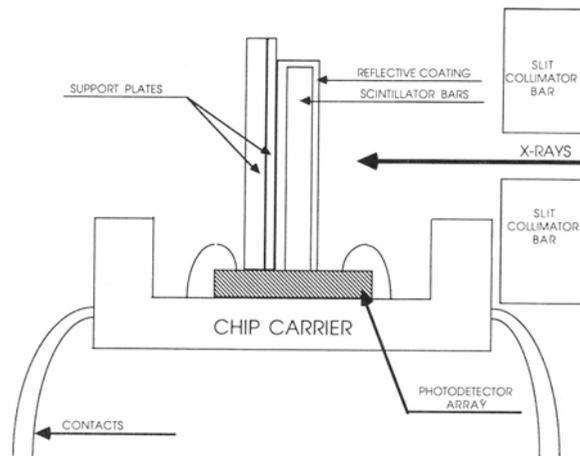

Figure 6.   X-ray detector array, end view.

Since the scintillator blocks are embedded in a reflective medium and directly bonded to an integrated circuit photodetector, the device described here is both rugged and compact. Amplifiers onboard the photodetector chip convert the output signals to a multiplexed, simply cabled form. This results in a small, robust detector that can be easily manipulated into small openings for convenient inspection of oddly shaped objects.

Small assemblies of composite materials are best inspected with soft x rays generated by tubes excited by less than 100 kVp. The design described here is especially sensitive to these easily absorbed x rays, because the scintillator blocks are shielded by only a thin layer of optical reflector.

SCINTILLATOR MATERIAL

These detectors are fabricated from a new ceramic scintillator material developed by GE. This engineered material, commercially known as HiLight, combines favorable x-ray performance with physical properties conducive to the micromachining required by our detector design.

The ceramic scintillator is a transparent, cubic, polycrystalline material composed primarily of a unique combination of rare-earth oxides that give rise to high x-ray stopping power, high scintillation-efficiency, uniform light output, and high resistance to radiation damage. The scintillator material is prepared from high purity ($>99.99\%$), fine ($<1\,\mu$m), chemically homogeneous powders that are pressed into powder compacts. A special high-temperature sintering process transforms the porous compacts into fully dense, optically transparent ceramics.

The ceramic scintillator has many desirable properties. It has a relatively high intrinsic density of ~ 6 g/cm^3 with an x-ray absorption coefficient of ~ 25 cm^{-1} for 70 keV x rays. Under comparable conditions the ceramic scintillator exhibits a scintillation efficiency about three times that of single-crystal cadmium tungstate, a luminescent afterflow that is 1/9 that of CsI:Tl scintillators, and a radiation damage resistance that is 20 to 100 times higher than that of CdWO$_4$ and CsI:Tl scintillators. The ceramic scintillator emits in that part of the visible region where the photodiodes exhibit excellent sensitivity.

Finally, the fired ceramic can be mechanically worked to tight tolerances and smooth surface finishes that are required for high-resolution detectors. The excellent machinability of the ceramic scintillator is due primarily to its small average grain size of $\sim 6\,\mu$m.

IMAGING PERFORMANCE

Two images that illustrate the current performance of this imaging system are included in this report. Both were made with the tube driven to 16 W of anode power at 80 kVp.

Figure 7 is a digital radiography image of a lead line pair gauge illustrating the basic system resolution. The left side of the photograph is a computer display of the data showing the five radial stripes of the gauge radiating down from an alignment hole at the top of the object. The gauge was scanned horizontally past the detector. Each horizontal line in the image represents the numerical average of eight consecutive detector readings at 80-ms intervals. The right side of the image contains a vector digitally clipped from the image at the 16 line pair/mm position on the gauge.
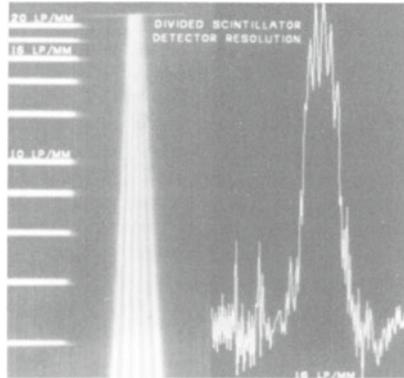
436

Figure 7.   Digital radiography image of line pair gauge.

Figure 8 is a computed tomography slice taken through the eye of a fine sewing nee-
dle. It was reconstructed from 720 evenly spaced projections through the 20-mil needle
using a STAR array processor. For both Figures 7 and 8 the response of individual
detector elements were normalized between 1.0, equaling their signal from the unat-
tenuated x-ray beam, and 0.0 as their no x-ray output. The images are displayed with full
grey scale expansion and no subsequent image processing to clean up artifacts.

ACKNOWLEDGMENTS

Figure 8.   CT slice through the eye of a needle.

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