INTRODUCTION

Though well established in medicine as a technique of inspection, x-ray computed tomography (CT) is just beginning to have an impact on nondestructive evaluation. A recent review [1] emphasizes the potential of the technology. An aspect which is apparent from the description of most current systems is that they are being applied to inspection of objects which tend to be locally uniform in the axial direction, e.g., electrical and telephone poles, trees, and rocket motors. The CT systems used are basically adaptations of medical CT technology, and reconstruct slices of an object one at a time. Under the hypothesis that single- or few-slice CT examination will not suffice for certain NDE problems, we have been developing a tomographic system which performs direct three-dimensional reconstruction, in the sense that the transaxial slice is not the fundamental unit of reconstruction. This is accomplished by collecting a set of 2D projection images and performing a reconstruction directly into a 3D array, using an algorithm developed for the purpose. The result differs from a set of spaced conventional slices primarily in the respect that our spatial resolution in the axial direction is substantially as good as that in transaxial planes. Our developmental system is currently limited by beam energy and physical size to examination of small objects; because of its scale it has better absolute resolution (approx. 0.1 mm) than is generally found in CT equipment. The principles employed are quite general and, given appropriately energetic radiation and a 2D array of detectors, could be applied to the inspection of much larger objects. The system will be described, and example reconstructions will be used to illustrate salient aspects of the methodology.

SYSTEM CONCEPT

Our approach to 3D reconstruction is geometrically very simple and is illustrated schematically in Figure 1. The specimen to be inspected is situated between a point source of x rays and a two-dimensional x-ray detector. The specimen is rotated in steps relative to the remainder of the apparatus. (Equivalently, the apparatus could rotate about the specimen as is usual in medical CT.) We refer to the direction defined by the rotational axis as the vertical direction, and refer to planes perpendicular to the axis as horizontal planes. The particular horizontal plane which contains the x-ray source is called the midplane.
The 2D x-ray detector system is thought of as planar; if the actual physical realization of the detector is nonplanar, it can be mapped into a plane. Conventional 2D CT may be regarded as a special case of 3D reconstruction in which one uses only the line of the detector system which is formed by its intersection with the midplane and reconstructs that portion of the specimen which lies in the midplane.

![Diagram of 3D x-ray tomography](image)

Fig. 1. Geometry for 3D x-ray tomography.

Data acquisition involves collecting a projection image of the specimen at each angle of a set uniformly spaced in a full circle. As used here, the term "projection" means a line integral of the attenuation coefficient (frequently referred to as "density") along a ray originating at the source and terminating at some position on the detector. By measuring the transmission $T$ along such rays, we may estimate the projection integrals $P$. A 2D array of such integrals forms a (sampled) projection image. The approximations inherent in this procedure have been well discussed in the literature; see, for example, Ref. [2].

The reconstruction procedure places a restriction on the geometry of the specimen relative to the system. We define the reconstruction region as that portion of the specimen between a pair of arbitrarily located horizontal planes which bound the region to be reconstructed. These planes are typically, but not necessarily, located symmetrically about the midplane. The restriction is then that all rays from the source which pass through the reconstruction region must be included in the projection image. This is equivalent to the requirement for conventional 2D reconstruction that the shadow of the cross section of interest be fully contained in the data. Use of a different reconstruction method (e.g., an iterative algebraic method) might allow some relaxation of this restriction, but the time penalty which would accompany such a change could make it impractical for 3D reconstruction.
The geometry used here may be regarded as the prototype for a flexible inspection station which combines the capabilities of real-time fluoroscopy and computed tomography. The image processing and computational resources needed to perform 3D CT may be used to great advantage to process fluoroscopic images.

RECONSTRUCTION ALGORITHM

We here briefly describe our convolution-backprojection algorithm, which previously has been discussed in detail [3]. As is the case in 2D convolution-backprojection algorithms, each angle of data generates a contribution to each reconstruction point, and the contributions from all angles are simply added together. (This contrasts with reconstruction methods in which the data from different angles interact or in which the data are treated iteratively.) Each raw image of the specimen is subjected to background correction and intensity normalization to produce a 2D array of transmission values. Estimates of the required projection integrals are then derived from the formula \( P = \log(1/T) \). [Though we do not consider it here, a simple correction for the phenomenon of beam hardening (see, for example, Ref. 2) may be incorporated by somewhat altering this relationship.] The projection image is subjected to a geometrical weighting and is then filtered along horizontal lines with a chosen convolution kernel. This convolution step is essentially identical to that performed in the commonly used 2D fan-beam algorithm; because of this, most of the accumulated wisdom regarding choice of the kernel applies here also. The backprojection operation assigns a contribution to each desired point of reconstruction. First we compute the intersection of the detector plane with the line defined by the source and the reconstruction point; then the value of the weighted and convoluted projection image at that intersection is multiplied by a further geometrical weighting factor and added to other contributions for the same reconstruction point.

As discussed in Ref. 3, our 3D algorithm is not exact. However, the consequences of its approximate nature seem to be negligible in practical use. A comparison is provided by the fact that, in the midplane, the algorithm reduces precisely to the standard fan-beam algorithm: a qualitative comparison of horizontal reconstructed planes away from the midplane discloses that little significant difference exists.

No real time penalty is attached to 3D reconstruction; i.e., the time required per reconstructed point is comparable to that for the 2D fan beam. Of course, the number of points in a 3D region tends to be fairly large so that the overall reconstruction time can be considerable. In this regard, we note that time can be conserved by reconstructing only the region of interest. By saving the projection data, the procedure can later be rerun for a different region or for reconstruction points spaced differently.

The reconstruction procedure produces an estimate of the linear attenuation coefficient at each point of reconstruction. Typically these are arranged on a 3D lattice. Planes of this lattice may be displayed as slices. As suggested in Fig. 2, various orientations may be chosen, most commonly horizontal (transaxial) and vertical slices. Horizontal slices are analogous to those of conventional CT; vertical slices portray the specimen in the same orientation as that of a live or digitized transmission image, often with much greater detail. Shaded 3D display is also possible; we have found the method described by Farrell et al. [4] to be quite useful for bringing out the architecture of the specimen. Various graphics and image processing techniques may also be brought to bear on such output images.
Fig. 2. Typical slices available from a 3D reconstruction. The orientation of the vertical slices depends on the zero of the angular coordinate used in reconstruction; by changing this value and repeating the reconstruction, the entire reconstruction mesh may be rotated.

PRESENT IMPLEMENTATION

We now discuss our implementation of the ideas presented above. It must be emphasized that this is a developmental system, designed for inspection of small objects, and that many improvements are possible. A block diagram of the system is shown in Figure 3.

Fig. 3. Block diagram of the 3D x-ray tomography system.
A Magnaflux microfocus x-ray source approximates a point source. Its minimum focal spot size of 0.05 mm is substantially larger than that of currently available sources and is a principal contributor to the resolution limitations of the system. The specimen is placed as close as possible to the source to effect a large geometric magnification, and is positioned and rotated by Aerotech translational and rotational stages under computer control. The detector system consists of an x-ray image intensifier (Precise Optics), a vidicon (Hamamatsu C1000-01), and a video digitizer (Quantex DS-12 or Grinnell GMR 270). Digitized images are summed to reduce the noise content and are transferred to a DEC VAX 11/730 computer to be processed. The VAX controls the data acquisition and reconstruction procedure, communicating with the Aerotech and the Quantex by means of an IEEE-488 instrument bus; the Grinnell, on the other hand, is a peripheral on the VAX system bus. A Floating Point Systems AP-120B array processor is also on this bus and performs the computationally intensive portions of the reconstruction procedure. An attached disk system augments the array processor memory, which is far smaller than that required for most reconstructions.

A considerable degree of parallelism exists in the system, in the sense that image digitization and summation, image correction, reconstruction, and buffered array-processor disk operations take place simultaneously.

Digitization of image intensifier output is an efficient means of producing a large 2D array of image samples (each digitized point is such a sample). However, several corrections are required to make such data sufficiently quantitative for our purposes. For example, beam intensity fluctuations are handled by using portions of the field (away from the shadow of the specimen) to monitor the intensity for each collected image. Angular nonuniformities in the x-ray output as well as sensitivity nonuniformities of the intensifier and camera are corrected by means of reference images (beam completely blocked and completely open) which enable a pixel-by-pixel estimate of transmission to be made.

Most conventional scanners utilize a discrete detector to obtain each requisite projection sample. The detector acceptance in the vertical direction is often larger than the spacing between detectors in the horizontal direction. The increased acceptance improves the signal/noise ratio but degrades resolution in the vertical direction. This is unimportant if the structure of the object varies slowly in the vicinity of the slice being imaged (as tends to be the case for the objects such as mentioned in the Introduction) but is detrimental in other cases. By its nature, our sample spacing and resolution is comparable in the horizontal and vertical directions.

Unlike that of a system of discrete detectors, the resolution of an image-intensifier-based detector system is not primarily determined by the spacing between the samples. The resolution is limited primarily 1) by the source spot size, 2) by blurring which occurs in the image intensifier and camera, and 3) by integration of the video signal during the analog-to-digital conversion which takes place in the video digitizer. These processes act as a spatial low-pass filter on the transmission image prior to sampling. To minimize aliasing, we generally take samples more closely spaced than the resolution width. In this respect, our system should be less susceptible to the effects of aliasing (e.g., the "partial volume effect") than systems which employ discrete detectors and hence have no mechanism for limiting the spatial frequency content which is presented to them, other than their finite size.
In order to achieve resolution in the 0.1 mm range, it is necessary to determine precisely the effective position of the rotational axis, because any error in this gives rise to a comparable blurring of the reconstruction. Likewise, the orientation of the rotational axis relative to the image plane must be known. By performing reconstructions of fine wires, we detect and correct small positioning or orientation errors.

**SYSTEM PERFORMANCE**

We state resolution as the diameter of the half-value contour of the reconstruction of a 3D delta (impulse) function. The resolution of our system, inferred from fine-wire reconstructions such as that in Figure 4, is typically in the range 0.100-0.125 mm.

![Fig. 4. Midplane reconstruction of a 0.1 mm diameter tungsten wire. In the reconstruction on the left the reconstruction points are on centers 0.05 mm apart. The reconstruction on the right employed the same projection data but was performed with a finer mesh (0.01 mm) and has been thresholded at a density value half that at the peak. The wire was located approximately 1.5 mm from the rotational axis; as long as the position of the rotational axis is sufficiently well known, the reconstruction does not depend significantly on the actual position. The effective sample spacing, referred to the specimen, i.e., the actual sample spacing divided by the geometric magnification (5.3 in this case) was approximately 0.04 mm. Projections were taken at 131 equally spaced angles. The reconstructions shown below were taken under the same conditions.

During acquisition of the projection data, we typically sum images at each angle until the standard deviation of the transmission is in the range of 0.01 for transmissions near unity. This, however, does not directly imply the sensitivity of the system to changes in density. Hence, we digress somewhat to discuss the relationship between the signal/noise (S/N) of projection images and that of a reconstruction based upon them. This is of particular importance for NDE applications, where the imaging of internal structure may be secondary to simply detecting some feature of interest, such as a flaw.

We consider the simple case of a cylindrical flaw in an otherwise uniform cylinder. Since all conclusions can be shown to be similar for three dimensions, we limit consideration to the midplane, i.e., to 2D reconstruction.

We first note that detection of a flaw by an imaging process depends largely on whether the contrast induced by the flaw is large enough compared to the noise intrinsic to the image. For example, in a line
plot with a given noise level, a rectangular bump is easily seen if its level is several standard deviations from the background, whereas a bump equal to one standard deviation cannot be located with confidence.

The available contrast from a flaw as observed in a reconstruction image is greater than that in a transmission or projection image because the density change is observed directly, rather than appearing as a small change in an integrated value. However, the convolution operation of the reconstruction process amplifies the noise which exists in the projection data; this may be understood as a propagation of errors due to the summation of many terms of both signs. (In a sense, increased noise must be accepted as a condition for obtaining spatial information.) The simple calculation which follows compares these competing effects.

Let the respective diameters of the cylinder and the flaw be D and d, and let their densities be given by F and f. The available reconstruction contrast is thus (F-f) and the projection contrast is d(F-f). Further, assume that the uncertainty in the projection measurement is given by δP. (To avoid possible confusion, note that this is roughly the uncertainty in the projection measurement at a single angle divided by the square root of the number of angles.) Denote the uncertainty in the corresponding density measurement by δF. The quantity

\[ R = \left( \frac{(F-f)}{6F} \right) \left[ \frac{d(F-f)}{\delta P} \right] \]

then expresses the appropriate comparison of the reconstruction and projection contrasts to their respective uncertainties. For a ray along a diameter, the relationship between δF and δP is found to be

\[ \frac{\delta F}{F} = 0.707 \frac{D}{S} \left( \frac{\delta P}{P} \right) , \]

where the numerical factor depends on the convolution kernel and S is the spacing between detector samples scaled to the specimen position. Using this relation, we find the simple form

\[ R = 1.414 \frac{S}{d} , \]

which is independent of the overall diameter and of the actual difference of the density of the flaw from that of the background cylinder. This result suggests that purely from a noise standpoint, tomography offers no advantage for large flaws (S/d << 1). In the case of very small flaws, the physical flaw size d must be replaced by a value comparable to the resolution. Then if S is about half of the resolution, as it should be for proper sampling, the value of R is of order unity. Under these conditions the flaw should be much more easily discerned in the reconstruction, since the flaw response is superposed on the relatively flat background density from the cylinder, rather than on the shape dependent background which is characteristic of a projection image.

Let us consider a specific application of the relation between the noise in reconstruction and the noise in the underlying projection images. Consider a uniform cylinder whose density F is such that \( T = \exp(-P) \approx \exp(-DF) = 0.5 \) for a ray through the center. If the uncertainty in T is 0.01 at T=1, the fractional uncertainty in T at T=0.5 will be \( \delta T/T = 0.014 \) and, correspondingly, \( \delta P/P = 0.02 \) for a single angle. Thus if we employ 100 angles and have S=0.01D, we find

\[ \frac{\delta F}{F} = 0.707(100)(0.02/10) = 0.14 . \]

Improvements in this value may be achieved by increasing the signal/noise
at each angle or by employing more angles. Apart from the cost of increased computation time, the latter is preferred as it has the additional effect of reducing the chance of artifacts caused by angular undersampling. As is well known, decreased noise may also be achieved by sacrificing resolution; here it may be done by smoothing the projection data (most efficiently done by broadening the convolution kernel) or by smoothing the reconstruction itself.

EXAMPLES

As mentioned above, wire reconstructions as in Figure 4 are useful for determining the system resolution as well as for checking that the position of the rotational axis is known sufficiently well. Any imprecision in this position manifests itself as a blurring of the reconstruction; a large error would actually result in a point being reconstructed as a ring. In setup of the system, it is useful to perform a wire reconstruction using only three angles. The backprojections from these angles should meet at a point, and any rotational axis position error is apparent by formation of a small triangle.

Figure 4 also illustrates that the same projection data may be reconstructed in different ways for different purposes. The coarse reconstruction was performed on our standard 0.05 mm mesh, while for detail the finer mesh is useful.

The left image in Figure 5 is a horizontal slice of a cylindrical plastic test object. The outer diameter is 10.3 mm and the inner diameter is 5.0 mm. The object itself is fairly uniform but has voids of various sizes as well as small regions of higher density. The right image shows the result of inserting an alumina thermocouple tube, of outer diameter 2.2 mm, at an arbitrary position in the center of the plastic cylinder.

Fig. 5. Horizontal midplane reconstructions of a plastic test object (left) and the same object with a ceramic thermocouple tube inserted. Much of the fine structure is real, as suggested by its reproducibility in these separate scans. Note the faint streaks in the lower left corner of the right image, caused by incomplete cancellation of backprojection from the ceramic material. The overall brightness increase in the center of this figure is due to the photographic reproduction process. The reconstruction points (and display pixels) are spaced 0.05 mm apart.

Faint streaks appear to radiate from the higher density ceramic material and would be reduced by use of more angles. Conversely, use of substantially fewer angles would cause additional streaks to originate
from the large density change at the inner wall of the cylinder. The holes in the thermocouple tube, though not particularly small (0.8 mm) were difficult to see in a fluoroscopic image because their small effect on the overall attenuation tends to get lost in the nonuniform background from the rest of the specimen. In reconstruction, they are quite clear. It is interesting to note that if fewer angles are employed, sharp features such as these holes tend to remain clear long after the overall reconstruction has been riddled with streak artifacts.

In Figure 6 we display horizontal slices of portions of the ceramic material of a catalytic converter before (left image) and after application of the heavy-metal containing washcoat. The attenuation coefficient of the washcoat is much greater than that of the ceramic material and hence dominates the reconstruction, even though the washcoat thickness is much smaller than the system resolution. The uniformity of washcoat application would appear to be easily judged from such reconstructions.

![Image of horizontal reconstructed slices of a portion of a ceramic catalytic converter monolith (left) and the same material after application of a washcoat.](image)

In each of the examples discussed thus far, cross sections of the specimen perpendicular its "axial" direction tend to be nearly identical. By averaging over a range of vertical slices, improved signal/noise can be obtained. Whether achieved in this way or by increasing the vertical detector acceptance as in conventional scanners, such improvement carries as a tradeoff 1) the chance of missing details which change rapidly in the vertical direction, such as the fine structure in Figure 5, and 2) increased blurring if the natural axis of the specimen is not precisely in the vertical direction.

Figure 7 shows a horizontal slice of a piece of glass-reinforced composite material. The various plies of the composite are clearly visible. A slice in one of the two directions orthogonal to this would likewise cut across the layers. The remaining direction is parallel to the layers and offers a quite different view, as shown in Figure 8; the selection of slices shown encompass about half the thickness of the specimen. The fiber bundles which make up the randomly oriented layers do not lie neatly in planes, resulting in a degree of
fuzziness in a planar slice. By combining many adjacent closely spaced slices into a 3D representation, as in Figure 9, a good idea of the internal architecture of the specimen can be obtained.

Fig. 7. Horizontal slice of a piece (cross section 3.71 mm by 7.04 mm) of glass-reinforced composite material. Only the bundles of glass fibers are visible in the linear gray scale used here.

Fig. 8. A sequence of vertical slices approximately parallel to the layers of the composite material specimen. The reconstruction region is 7.95 mm high.
SUMMARY AND CONCLUSIONS

We have discussed the concept and implementation of a true 3D tomography system for nondestructive evaluation. We have shown several examples of reconstructions taken with the system. The system has been shown to be useful in applications beyond those of industrial NDE, e.g., biomedical applications involving resolution requirements which cannot be met by conventional scanners. The concept can easily be scaled to larger dimensions. The computational requirements for full 3D reconstruction roughly scale with the number of reconstruction points and are in proportion to those of conventional reconstruction. Further, the reconstruction scheme lends itself to parallel computation and hence is well positioned for advances in this field.

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REFERENCES

DISCUSSION

Mr. Basil A. Barna (Idaho National Labs): In one of the images, there were some dark bands?

Mr. Feldkamp: As I noted, those are reconstruction artifacts, which here radiate from the small, higher density region of the cross section. The latter, in effect, contains a high proportion of both spatial and angular high-frequency components. To eliminate such artifacts from this reconstruction would require more angles and possibly a somewhat smaller effective detector sample spacing. Such additional effort would be justified if, for example, a search for a feature of very low contrast would be hampered by such artifacts.

Mr. Thomas Derkacs (TRW): The alternative to what you are doing, as you mentioned, would be to take a planar tomograph and construct the three-dimensional object one at a time. Do you have any feel yet for the difference in terms of the time to construct the image between doing that and doing the whole three-dimensional calculation?

Mr. Feldkamp: On the basis of time per point reconstructed, we pay almost no penalty for performing the calculation in three dimensions, the only additional time amounting to overhead which becomes fractionally insignificant as the vertical extent of the reconstruction amounts to twenty or more slices. From the standpoint of data gathering, the cone-beam geometry used here is more efficient than the slice-by-slice approach, just as the fan-beam method is more efficient than the parallel-beam approach.

Adapting the calculation to specialized hardware may be more difficult than in the fan-beam algorithm, if the approach of storing interpolation coefficients is followed, because of the multitude of such coefficients required in three-dimensions.