Quantitative Microtomography

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Introduction

In the late 1970s and early 1980s, Elliott was seeking to quantify and map mineral concentration in human tooth samples, an essential element in the study of dental pathology and treatment. In the first instance, thin tooth sections were imaged with contact microradiography. Using an aluminium step wedge, the density of the X-ray film image could be related to the mineral concentration in the sample. To obtain better accuracy, the film was replaced with a 2D scanning microradiography system whereby the thin sample was slowly scanned through a pencil X-ray beam and the attenuated beam recorded with an energy discriminating photon counting system, configured to count photons in the X-ray target characteristic emission lines. The remaining uncertainty was the sample thickness, which was difficult to ascertain precisely because of difficulties in cutting and dependence on the level of hydration. This led to the development of a first generation microtomography system (Elliott and Dover, 1982), using the same photon counting system, that gave accurate measurements of the X-ray linear attenuation coefficient (relating to mineral concentration) independent of sample geometry. Thus began the history of this group's development of X-ray microtomography as a means primarily of mapping the X-ray linear attenuation coefficient (LAC), as opposed to concurrent commercial developments where the imaging aspect of microtomography was of greatest interest.

From 1st to 4th generation?

Because of its energy selectivity, the first generation system was in many ways the gold standard for mapping the LAC, but it was very slow. Data for a single slice of 128 X 128 pixels took 12 hours to collect. To obtain the same quality data in a 512 X 512 X 512 volume would take around 45 years! There was thus a need to follow the trend of other developers and use area X-ray detectors to increase data acquisition speed. At this time, medical scanners were using 4th generation designs with a fixed ring of detectors and rotating X-ray source. This meant that each projection was effectively recorded by a single detector element, thus eliminating ring artifacts in the reconstructed image. Since it was not economically (and maybe physically) possible to create a detector ring small enough for microtomography, we used a linearly moving CCD camera with time-delay integration (TDI) readout to achieve the same advantage, though at the cost of increased data acquisition time relative to conventional readout (but still orders of magnitude faster than the first generation system).

In search of perfection

The switch to the TDI system (later dubbed MuCat; Davis and Elliott 2003) brought a huge increase in speed, but since the detector was no longer energy selective, measurements of the LAC were less accurate. Furthermore, the area detector was more exposed to scattered radiation than the single collimated photon detector, bringing a further reduction in accuracy. Over the years, improvements have been made in accuracy by means of scatter reduction and beam hardening correction.

Scatter reduction

Scatter generally increases with increasing sample size. In the TDI system, the camera is traversed across the X-ray shadow. By use of a lever arm mechanism, we have added a moving aperture in front of the source, collimating the beam so that it tracks the moving camera. This illuminates only the part of the specimen that is necessary and thus reduces scatter. Our current MuCat 2 scanner has a 4K X 4K detector which is ultimately binned to 1K X 1K pixels. With TDI motion, this can produce an image of up to 2,6K X 1K pixels (limited in width by the beam angle), but the whole 2.6K width is not illuminated at the same time.

Beam Hardening Correction

A common approach to beam hardening correction is to adjust one or more parameters in a linearization process such that the reconstructed image has the same density in the middle as the outside. This is a good approach for many applications where it is known that the density should be the same inside and out. Teeth, for example, have a real radial gradient in mineral concentration and so this approach is not valid. We have used a pure aluminium step wedge to derive a polynomial linearization curve. This works well provided that the attenuation vs energy characteristic of the sample is close to that of aluminium and provided that the maximum attenuation of the sample does not exceed that of the step wedge (polynomials are good for interpolation, but terrible for extrapolation). This curve converts attenuation values made with polychromatic radiation to equivalent attenuation with monochromatic radiation. Typically, for an X-ray (accelerating) voltage of 90 kV we will calibrate to 40 kV monochromatic radiation. Beam hardening correction has now been updated, using a modelling approach (Davis et al, 2008). This allows extrapolation beyond the maximum attenuation of the step wedge and also allows the linearization curve to be adjusted according to known specimen composition. To allow for slight changes in the spectrum between calibration scans, we include a pure aluminium wire with the sample where accurate LAC quantification is required. An adjustment is made to the recorded values based on the measured LAC of the wire.

Examples

The techniques above allow us to obtain XMT images with both high contrast ratio and high accuracy. Figure 1 illustrates the advantage of obtaining a high contrast ratio. This is from a study of dentine sealants in dentistry. A tooth sample was cut to expose the dentine and was treated with a sealer to which a small amount of a polymer contrast agent was added. In this figure, it can be seen that the contrast agent has penetrated along what is assumed to be “open dead tracts” (a response to dental decay) in the dentine.

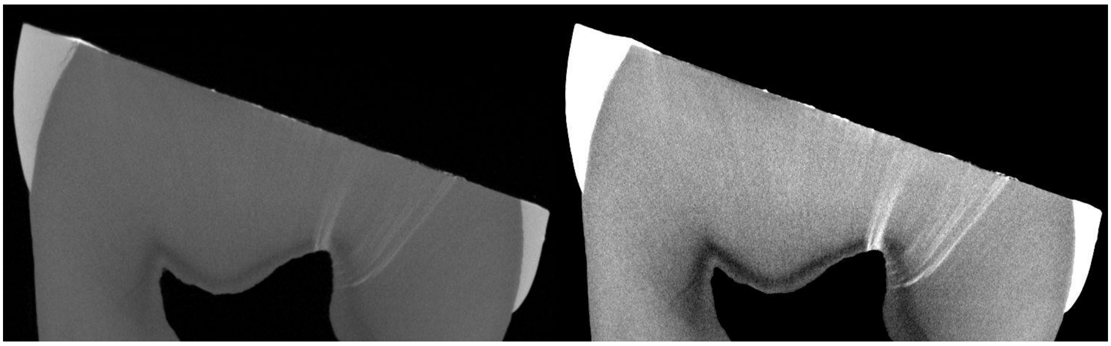

Figure 1: Sealant penetration in dentine: Normal contrast (left) and X4 contrast (right).

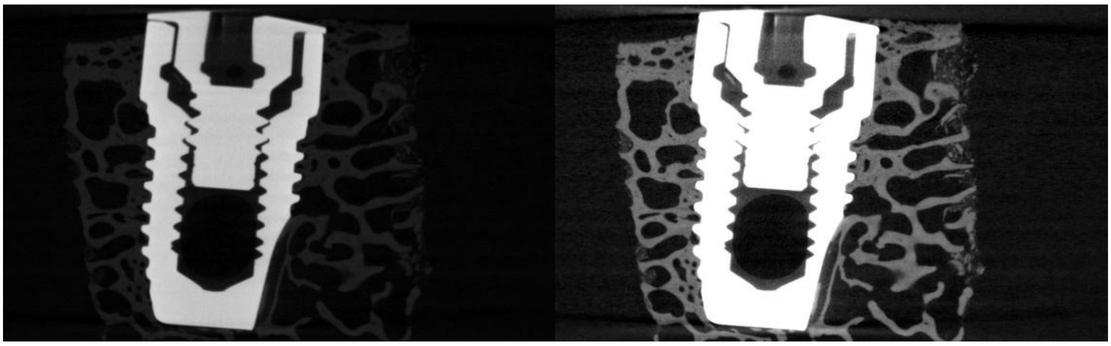

Figure 2: Titanium dental implant in sheep bone. Normal contrast (left) and X4 contrast (right).

A further study illustrating the use of TDI imaging with the modelling approach to beam hardening correction is shown in Figure 2. This shows a scan of a titanium dental implant in a sheep bone (in vitro). Titanium is approximately 7 times more attenuating than bone over the energy range used, hence the bone is barely visible at normal contrast. Previously, beam hardening artefacts from such scans caused severe errors in analysis of the bone, especially in the vicinity of the implant (which is the most important region for the research). Using the modelling approach, with adjustment for titanium, bone could be seen right up to the implant interface with little sign of artefacts.

Conclusion

Quantitative microtomography is problematic with conventional X-ray sources because of the effects of polychromatic X-rays and scattering. Some experimenters have even concluded that acceptable results can only be obtained with a synchrotron system. We have demonstrated that this is not the case. Improvements in accuracy and image quality have come slowly over the years through incremental development of scanning and processing techniques and we anticipate further improvements in the coming years.

References

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