Laser driven x-ray imaging of human trabecular bone

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Introduction
There is a pressing need in both histological and diagnostic medical imaging to be able to examine the structure of human bone on the micron level\textsuperscript{1-3}. Trabecular bone is the spongy and porous bone layer responsible for most internal metabolic activity and red blood cell production. Due to the high rate of bone calcium turnover there, trabecular bone is affected severely by osteodegenerative diseases such as osteoporosis\textsuperscript{4}, leading to the appearance of microcracks\textsuperscript{5} and thinning or broken trabeculae. Detecting these characteristic early signs of osteoporosis in the bone would help in diagnosis and state of the art clinical drug trials\textsuperscript{6}. The complex, small scale structure and high atomic density of trabecular bone places stringent demands on any proposed x-ray source – it should be high energy, have a small source size and produce a relatively high flux of photons.

Microfocus x-ray sources are commonly used in clinical settings\textsuperscript{7}, producing K\alpha and bremsstrahlung x-rays by focussing an electron beam onto a high-Z metal anode. Unfortunately the achievable x-ray flux is limited by the melting point of the anode – focussing the electron beam to reduce the source size heats the anode. If the anode is to stay solid the electron flux, and hence x-ray flux, must be reduced\textsuperscript{8}.

The laser-wakefield acceleration (LWFA) technique\textsuperscript{9-11} relies instead on a plasma to accelerate electrons longitudinally and transversely, acting as both the accelerating and radiating medium. As a consequence it is possible to generate tightly focussed and high current electron beams without damaging any ‘target’ material, and the dependence of x-ray flux on source size is lifted. Here we demonstrate the use of a laser-wakefield driven x-ray source to take high-resolution absorption contrast radiographs of a human femoral trabecular bone sample in single shots.

Experimental Setup
The experiment described here took place at the Astra-Gemini beamline\textsuperscript{13}, which delivered pulse energies of 11.4 ± 0.4 J in 45 fs at 800 nm. The 150 mm diameter linearly-polarised Gemini beam was focussed with a 3 m focal length (f/20) parabolic mirror into a slightly elliptical focal spot of 25 × 32 μm FWHM. The peak laser intensity at vacuum focus was then (1.80 ± 0.6) × 10\textsuperscript{19} W cm\textsuperscript{-2}. The line- of-sight thickness of the bone at a particular pixel. The variation of D over the CCD defines a projected image of the sample, whereas the (strong) variation of μ with photon energy defines the contrast of the image. Taking both of these into account, if the photon distribution across the sample is assumed to be Poisson-like, the signal to noise ratio (SNR) can be calculated to scale like\textsuperscript{8}.

Choosing Beam Parameters
This experiment considers the formation of images through absorption contrast, where the x-ray transmission of the bone (assuming a constant density throughout) varies as \( e^{-\mu D} \). Here \( \mu \) is the mass attenuation coefficient of the bone, and \( D \) is the line-of-sight thickness of the bone at a particular pixel. The variation of \( D \) over the CCD defines a projected image of the sample, whereas the (strong) variation of \( \mu \) with photon energy defines the contrast of the image. Taking both of these into account, if the photon distribution across the sample is assumed to be Poisson-like, the signal to noise ratio (SNR) can be calculated to scale like\textsuperscript{8}.

![Experimental layout within the Astra Gemini vacuum chamber](image-url)
where $n$ is the number of photons per pixel – the relative changes to SNR with sample thickness are plotted in Figure 2. For low/high photon energies the sample becomes opaque/transparent and absorption contrast signal is lost.

\[ \text{SNR} \propto \sqrt{\frac{n \mu^2}{2 k E_0}} \]

Maximising this expression for a given photon flux we find an optimal $\mu D^{-1}$. As mentioned above, $\mu$ is strongly dependent on photon energy, approximately inversely proportionally, so picking an optimal $\mu$ is equivalent to using photons of an optimal energy. For our bone sample, a 7 mm diameter cylinder, the simple expression above suggests image contrast will be maximised at a photon energy of approximately 30 keV.

When optimising the source in the experiment we focussed on the quality of the radiograph, achieved with a spectrum of critical energy $E_c = 36$ keV (the energy above which half of the energy is radiated in the synchrotron spectrum) – see Figure 3. Though the above estimate of photon energy is a simple one based on a monoenergetic x-ray beam, it nevertheless helps indicate the useful energy range of a broadband beam for a given sample.

This radiograph-centric optimisation process places less importance on perfecting the electron spectrum, reflected in Figure 3. Though it was possible to produce electron beams at energies above 1 GeV, $E_c$ scales with the square of electron energy, and at this point the sample would be too transparent. The beams typically consisted of a narrow peak superimposed on a broad low energy tail. The lower energy electrons likely had little impact on the radiograph as very few would be transmitted through the sample. Over the course of the imaging runs the electron and x-ray beams remained reasonably stable, as has been observed in previous experiments using gas cells rather than gas jets as a plasma target.

**Imaging Results**

After x-ray source optimisation it was possible to record many hundreds of high quality radiographs over several hours, each recorded on a single shot. One radiograph is shown in Figure 4, where the hexagonal background is a feature of the fibre-coupling to the CCD of the camera. Individual trabeculae are clearly visible throughout, as are fine features at the edge of the sample. Limited by the physical size of the CCD and the requirement that the whole sample be visible, the effective pixel size is 4.8 $\mu$m corresponding to a geometric magnification of 2.8. Though difficult to estimate without an on-shot resolution target, examination of the edge transfer function of sharp features in the radiographs suggests a 2D resolution of approximately 40 $\mu$m. This is well above the intrinsic source size of the LWFA ($\geq 1$ $\mu$m) and the effective pixel size, suggesting there are other factors at play affecting the final resolution.

**References**


