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Double Grating Interferometry in a Commercial Micro Computed Tomography System for Biomedical Imaging

Griffin Rodgers[1], Anna Khimchenko[1,2,3], Hans Deyhle[1], Joachim Schulz[4], Bert Müller[1,*] and Georg Schulz[1]

[1.] Biomaterials Science Center, DBE, University of Basel, Allschwil, Switzerland.
[2.] Department of Dermatology, Harvard Medical School, Boston, USA.
[3.] Wellman Center for Photomedicine, Massachusetts General Hospital, Boston, USA.
[4.] Microworks GmbH, Eggenstein-Leopoldshafen, Germany.
* Corresponding author, bert.mueller@unibas.ch

X-ray grating interferometry is a phase-contrast imaging technique that provides high contrast, local electron densities, and can be translated to laboratory and medical settings. This phase contrast modality is particularly suited for the visualization of physically soft tissues and specimens that contain both soft and hard tissue components with micro computed tomography (μCT) [1]. The integration of an X-ray double grating interferometer (XDGI) into a commercially available μCT system could provide access to this technique to a greater audience. However, commercial systems present many obstacles to such modifications, particularly due to their compact nature with respect to custom systems. In this work, we discuss the design of the XDGI setup in a commercial μCT system and demonstrate its operation [2].

An X-ray double grating interferometer was incorporated into a nanotom® m commercial laboratory μCT machine. This system does not need a source grating thanks to its microfocus tube with a range of focal spot sizes between 0.9 and 2.3 μm, as measured from radiographs of a Siemens star pattern. The system allows for a maximum source-detector distance of nearly 60 cm, limiting the ability optimize angular sensitivity by going to higher Talbot orders. The interferometer was designed for operation in a symmetric setup that uses this full distance, with source-G1 and G1-G2 distances both set to 29.4 cm. The interferometer consists of a G1 with 7 μm period and gold lines 5.8 μm high, designed to achieve a π phase shift for operation at 30 keV and a first fractional Talbot distance at 29.4 cm in the specified geometry. The absorption grating G2 has 7 μm period and gold lines 70 μm high.

A part of a human knee joint was selected for the phase-contrast μCT, because it contains both hard and soft tissues. For this reason, the knee of an 87-year-old female was fixed in 10 % formalin and a cylinder 5.4 mm in diameter containing bone and cartilage was extracted from the femoral-tibial joint.

Measurements of visibility and counts were made over a range of acceleration voltages for each focal spot size. The angular sensitivity [3] was optimized for a spot size of 2.0 μm and an acceleration voltage of 42 kVp. Figure 1a shows the visibility map for our setup, which achieved a mean visibility of 25 % over a 3 cm × 6 cm region.

Phase- and absorption-contrast μCT measurements of the knee joint specimen were performed. Figure 1 shows a cross section in phase-contrast (b), simultaneously acquired absorption-contrast (c), and optimized absorption-contrast data from the nanotom® m (d). The phase measurement shows sufficient contrast to resolve the cartilage, unlike both absorption measurements. The contrast-to-noise ratio (CNR) of cartilage to formalin was measured as 1.094 ± 0.152 for phase channel, 0.073 ± 0.007 for the absorption channel, and 0.287 ± 0.003 for the optimized conventional absorption measurement.

A limitation of the current setup is the low count rate due to high X-ray absorption from an Al film in

front of the detector, a common feature of medical panel detectors that usually operate at higher photon energies. This leads to exposure times of 12 s per phase step and total scan times around twelve hours. Such long scan times create issues with setup stability and preclude high throughput investigations.

While it was possible to visualize cartilage, the CNR of around one indicates the electron density difference between formalin and cartilage, measured to be $(2.20 \pm 0.30)\ 10^{28}$ electrons/m^3, is at the sensitivity limit of the current setup. For context, variations of electron density within a formalin fixed human brain are on the order of $0.05\ 10^{28}$ electrons/m^3 [4]. Therefore, an optimized interferometer setup is desirable for imaging samples composed of entirely soft tissues. We are currently designing this setup with the help of numerical simulations incorporating the energy spectrum of our instrument, finite source size, cone beam geometry, and the non-negligible absorption of the phase grating. We must additionally consider the sensitivity as described [3] and the influence of the source size and geometry on the visibility [5]. Due to the geometrical constraints of the commercial system, we plan to design two complementary setups, i.e. a high-energy setup for directional dark-field measurements and a low-energy setup for phase-contrast measurements. Despite count rate issues, phase sensitivity is generally significantly improved at lower energies thanks to the quadratic energy dependence [6].

References:

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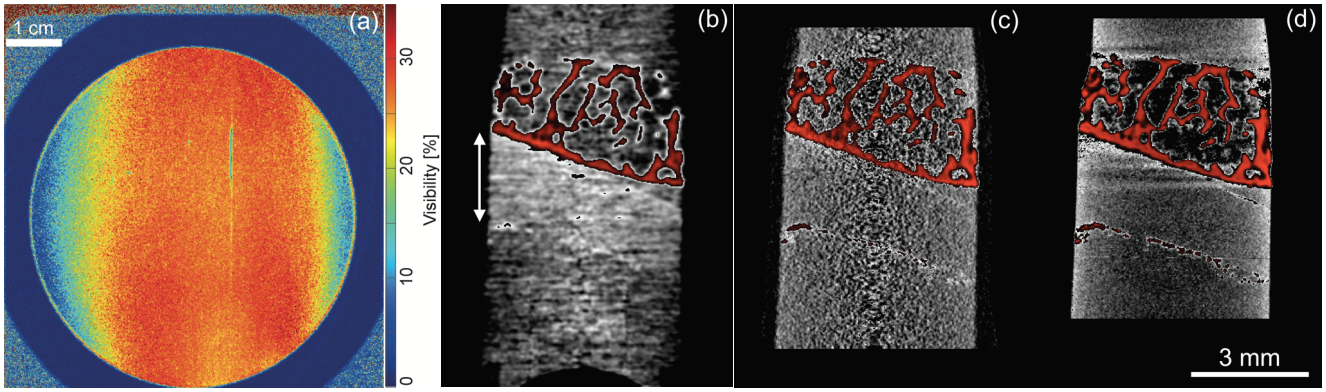

Figure. 1. Visibility map of the XDGI setup (a) and cross-sections of a human knee joint obtained with a commercial laboratory µCT-system equipped with the described XDGI, for phase contrast (b). Simultaneously acquired absorption contrast (c) and optimized, dedicated absorption contrast measurements (d) in the same system show reduced contrast in cartilage (white arrow in b) [2].