Artefact Suppression Algorithm in X-Ray Phase-Contrast Computed Tomography

Phase-contrast X-ray CT using refraction angle analyzer crystal can delineate biological soft tissues with a much higher contrast than conventional absorption-contrast CT. However, if the tissue specimen includes bones or calcifications, they generate extensive artefacts due to the large difference in the refractive index, leading to suboptimal visualization. In order to improve the visualization of an object composed of hard and soft tissues, we proposed an iterative reconstruction algorithm with artefact suppression and demonstrated its efficacy using data acquired on a murine model of rheumatoid arthritis.

X-ray phase-contrast imaging (XPCI) explores an alternative mechanism of interaction between the X-ray wave and tissue, namely phase alteration or bending of X-rays due to electron clouds of various materials. Even though different types of soft tissues have very similar attenuation, they have widely varying refractive index which is responsible for markedly different phase alterations imparted by different types of soft tissues. This can be picked up by a variety of phase sensitive methods such as X-ray interferometry, in-line holography, angle analyzer in crystal optics and grating optics. As a result, XPCI can provide high image contrast for soft tissue structures.

A key technical limitation of analyzer-based phase imaging systems is that they cannot accurately reconstruct a tomographic slice when the X-ray propagation suffers a large change in direction. The reason for this limitation is as follows: Phase-contrast X-ray tomography, just as in conventional attenuation-based tomographic reconstructions, requires that the intensity in the projection images represents a Radon transform of the differential refractive index map of the object being imaged. The intensity in the projection images varies with the beam deflection only for a limited range. Within this range, the propagation direction falls within the measurable region of the rocking curve of the analyzer in crystal optics. However, for a large change in direction at a tissue boundary, this condition breaks down because the direction deviates from the measurable region of the analyzer, resulting in a degenerate condition.

Soft-tissue structures are weakly refracting because the real part of their complex refractive index, the component that is responsible for phase alteration of the wave front, is very close to 1.0 for X-rays. Therefore, they cause only a small amount of bending in the X-ray wave front and the relationship between the intensity and angle of refraction is one-to-one. As a result, an accurate phase map can be obtained for soft tissues by applying the inverse Radon transform to the acquired projection images. On the other hand, dense structures such as bones and calcifications result in a large change in direction and violate the one-to-one condition: the amount of phase change can no longer be properly reproduced from the intensity in the acquired phase projection using the rocking curve. As a result, when standard methods for obtaining the inverse Radon transform are applied, the resulting phase image has extensive artefacts in the vicinity of dense structures. In their genesis, these artefacts are similar to the metal and beam-hardening artefacts seen in traditional attenuation-based X-ray computed tomography even though they tend to be more severe and have a different physical basis.

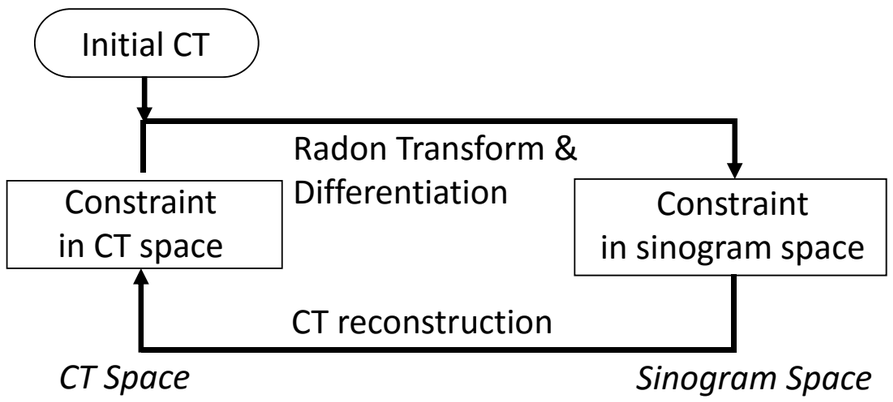

Figure 1: Block diagram of the artefact suppression algorithm.

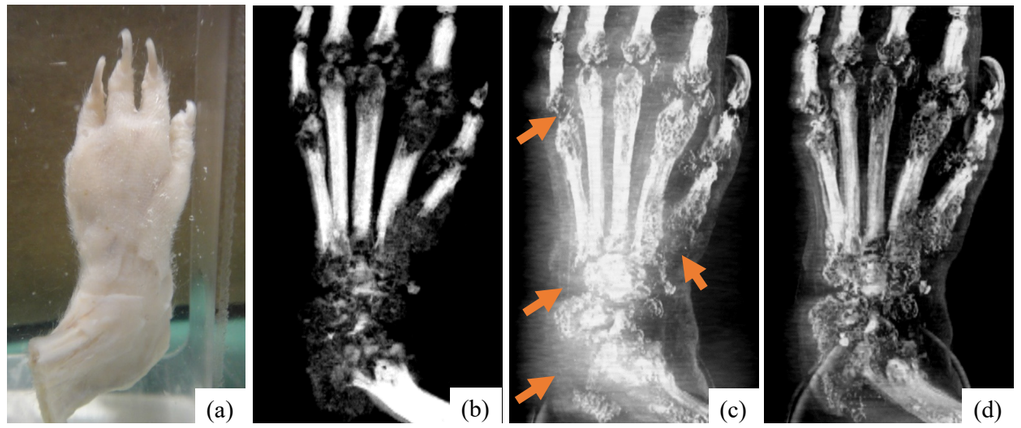

Figure 2: A photograph **(a)** and MIP images **(b)-(d)** of a rat model of human rheumatoid arthritis. Absorption **(b)**, phase contrast with bone artefact **(c)**, and phase contrast using the proposed algorithm **(d)**.

The absorption or attenuation image, on the other hand, depicts calcification well and this artefact is present only in the phase image. Since the attenuation image provides a map of the areas where the voxels responsible for phase aberrations reside, one can potentially suppress them using an iterative tomographic reconstruction algorithm. Based on this idea, we first proposed an iterative reconstruction algorithm that goes back and forth between a tomogram and its sinogram through the Radon transform and the inverse Radon transform while imposing a priori information in individual regions as shown by **Fig. 1** [1]. However, there is a drawback that this algorithm accumulates noise with each iterative step. These noises were suppressed by an improved reconstruction algorithm including a denoising step based on the total variation [2].

In order to evaluate the efficacy of the algorithm, we scanned a rat foot with a model of rheumatoid arthritis (**Fig. 2a**) using XPCI implemented via dark field imaging [3]. **Figures 2b-d** show maximum intensity projection (MIP) images reconstructed from the CT volume of the arthritic rat foot using absorption-contrast **b**, phase-contrast with bone artefact **c**, and phase-contrast with bone artefact removed using the proposed algorithm **d**, respectively. As expected, the absorption-contrast image (**Fig. 2b**) shows only the bony details and has very little contrast in the bone destroyed by arthritis. On the other hand, the phase-contrast images (**Fig. 2c** and **d**) show the detailed morphology of the affected joints and the surrounding soft tissues up to the skin surface. This figure also demonstrates the areas of signal loss due to large phase change in beam direction caused by dense structures. As can be seen in the regions marked by arrows in **Fig. 2c**, there is near-complete signal drop-out due to artefacts. The proposed algorithm dramatically improves the detail within these regions (**Fig. 2d**). In Ref. [2], a separate body of work that is not presented here, the proposed algorithm was also applied to human blood vessels and demonstrated the superior capability of soft tissue recovery as well.

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