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Evaluating the effects of detector angular misalignments on simulated computed tomography data

Massimiliano Ferrucci[a,b,*], Evelina Ametova[b], Simone Carmignato[c], Wim Dewulf[b]

[a] Engineering Measurement Division, National Physical Laboratory, Hampton Road, Teddington TW11 0LW, Middlesex, United Kingdom
[b] Department of Mechanical Engineering, KU Leuven, Celestijnenlaan 300—Box 2420, 3001 Leuven, Belgium
[c] Department of Management and Engineering, Università degli Studi di Padova, Stradella San Nicola 3, 36100 Vicenza, Italy

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ABSTRACT

The quality of dimensional measurements made by industrial X-ray computed tomography (CT) depends on a variety of influence factors in the measurement process. In this paper, the effects of angular misalignments of a flat-panel detector are investigated. First, a forward projection model is applied to evaluate distortions of the radiographic pixel coordinates assigned to X-ray intensities due to various detector rotation angles. Distortion maps are presented for a set of representative detector rotations and the sensitivity of image distortions to each rotation is discussed. It is shown from a simulation study that detector angular misalignments result in systematic errors of the reconstructed volume. The distortion model is inversely applied to generate correction maps that are used to correct the simulated radiographs from a misaligned detector. A new volume is reconstructed from the corrected radiographs and the new deviations are compared to the uncorrected results. The reduction of observed volumetric errors after radiographic correction validates the efficacy of the radiographic distortion model. Additionally, the output of this study can contribute to the development of a geometrical error model for volumetric measurements made by CT.

1. Introduction

In X-ray CT, the accurate reconstruction of the measurement volume is strongly dependent on the alignment of the system geometry [1]. For typical industrial CT systems, the geometry is defined by the relative position and orientation of the three main components [2], namely the X-ray source (particularly, the X-ray focal spot), rotation axis, and detector. To determine the sensitivity of the reconstructed volume to the alignment of these three components, the principles of X-ray CT are briefly revisited here.

The measurement volume is reconstructed by way of applying tomographic (slice-wise) reconstruction to a collection of radiographic images—or radiographs [3]. Typically, radiographs are taken in sequence as a test object is rotated on a stage. The information contained within each radiograph corresponds to the spatial distribution of attenuated X-rays incident on the plane of the detector. More specifically, the intensity registered by each pixel corresponds to the intensity of those X-rays that traverse the path from the X-ray source to the corresponding pixel. In the case of an object within the measurement volume, the registered intensity at each pixel will depend on the attenuation of X-rays by the object along the source-to-pixel path. The intensity at each pixel will also depend on scatter and other X-ray effects that are out of the scope of this paper, but are discussed elsewhere in the literature [3].

Radiographs are taken at multiple rotation positions, providing a denser sampling of X-ray attenuation trajectories through the measurement volume. The registered attenuation along all X-ray paths is used by the reconstruction algorithm to generate a three-dimensional distribution of relative attenuation values in the measurement volume. This volumetric model consists of three-dimensional pixels, ‘voxels’, with assigned grey values, which correspond to the relative attenuation at that voxel position. The position of each voxel is given by the three-dimensional coordinates of its centre. Subsequent processing of the volumetric data, such as segmentation and surface sampling, can be used to generate a three-dimensional point cloud; dimensional measurements can then be performed on the resulting coordinate points [1].

The accuracy of the extracted three-dimensional coordinates is dependent on the alignment of the system geometry and its stability during a scan. The grey value assigned to a voxel representing a particular volumetric space is calculated from

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[*] Corresponding author at: National Physical Laboratory, Hampton Road, Teddington TW11 0LW, Middlesex, United Kingdom. Tel.: +44 20 8943 7018.
E-mail address: massimiliano.ferrucci@npl.co.uk (M. Ferrucci).

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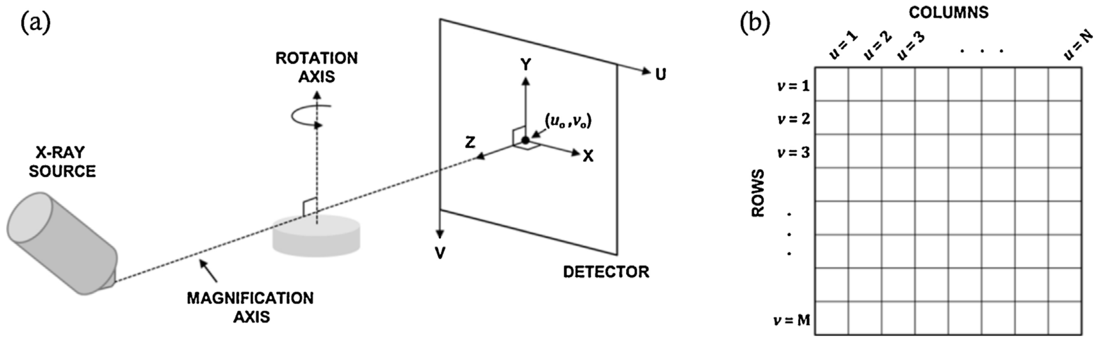

Fig. 1. (a) The ideal geometrical alignment of a typical industrial cone-beam X-ray CT system. (b) The pixel column indices increase rightward, while the pixel row indices increase downward. The pixel position $(u, v) = (1, 1)$ is located at the top left corner of the detector.

the set of X-ray trajectories through that volumetric space. Each trajectory through the measurement volume is determined from the assumed positions of source and pixel. Also, knowledge of the relative orientation of the measured object between radiographs is determined from the rotation of the stage, which is typically assumed to be stable [2]. Deviations in the system geometry from its assumed state will introduce errors in the radiographic pixel coordinates assigned to registered X-ray intensities. The propagation of these errors through the reconstruction algorithm will result in errors of the binning of grey values to individual voxels and, consequently, errors in dimensional measurements performed on the reconstructed volume.

In this paper, the effects of angular misalignments of a flat-panel detector on volumetric measurements made by CT are studied. First, the geometry of a typical cone-beam X-ray CT system is described. Then, a forward projection model [4] is adapted to generate radiographic distortion maps for various detector rotations. In practice, uncertainty in the input parameters would result in an uncertainty of the distortion maps. The scope of this study is to evaluate the effects of a misaligned detector; therefore, the input parameters for the model are assumed to be exactly known. It is shown by simulation that detector angular misalignments result in systematic dimensional errors of the reconstructed volume. The simulation study is briefly discussed and the volumetric errors are presented for various detector misalignments. The distortion model is then applied inversely to correct the radiographs from each simulated detector misalignment. A new, corrected volume is reconstructed with the corrected radiographs. The deviations from ideal geometry in the corrected volumes are compared to the deviations in the corresponding uncorrected volumes.

2. Instrument geometry

Fig. 1a summarizes the X-ray CT instrument geometry assumed for this paper. To begin, a right-handed global coordinate system is defined. The magnification axis, also the Z axis, is given by the line connecting the centre of the X-ray focal spot to the centre of the detector. The Y axis is parallel to the rotation axis of the object stage. The X axis is orthogonal to both the Y and Z axes, thus forming a Cartesian coordinate system. The origin is defined as the intersection of the ideal magnification axis and the detector plane. The positive Z direction is towards the X-ray source, while the positive Y direction is upwards (opposite the direction of gravity). The direction of the positive X axis follows the right-hand rule. The X-ray source-to-rotation axis distance (SRD) is given by the distance from the centre of the X-ray focal spot to the intersection of the rotation and magnification axes, while the source-to-detector distance (SDD) is given by the distance from the X-ray focal spot centre to the centre of the detector. Both SRD and SDD are positive values. The detector is positioned on the opposite side of the rotation stage from the source, thus SDD is larger than SRD. SRD is not an input parameter to the model but is mentioned here for reference.

The nominal alignment of the detector is as follows. The magnification axis (Z) is normal to the plane of the detector. The vertical axis of the detector (V) is antiparallel to the Y-axis, while the horizontal axis of the detector (U) is parallel to the X-axis. The flat panel detector consists of M by N pixels, where M is the number of rows and N is the number of columns (Fig. 1b). Ideally, the pixels are equally-sized and equally-spaced in the plane of the detector; the variables Δu and Δv correspond to the pixel width and height, respectively. The centre of each pixel in the detector is assigned column (u) and row (v) indices. The $(u, v = (1, 1)$ position is at the top left corner of the detector screen; the columns increase rightward (+X direction in the system coordinate frame), while the rows increase downward ($-Y$ direction in the system coordinate frame). The variables u_o and v_o are the pixel column and row coordinates, respectively, corresponding to the intersection of the magnification axis and the detector; this feature is also known as the principal point—a term commonly found in camera calibration for machine vision [5]. In the case of an ideally aligned detector, the principal point is located at the geometrical centre of the detector plane. Depending on the number of pixel rows and columns (even or odd), the centre of the detector can fall on a pixel or on the edge between adjacent pixels, *i.e.* u_o and v_o can be non-integer values.

Angular misalignments of the detector are described by three rotations: detector tilt θ about the X-axis (Fig. 2, left), detector slant φ about the Y-axis (Fig. 2, centre), and detector skew η about the Z-axis (Fig. 2, right). Tilt θ and slant φ are known as out-of-plane rotations, while skew η is an in-plane rotation. Detector rotations, in practice, are not constrained to occur about the central axes of the detector plane [6]; such rotations can be modelled as a combination of detector translation and rotation.

The effects of positional misalignments of the detector are not investigated in this paper to allow for in-depth analysis of angular misalignments. The principal point is therefore located at the detector centre. It should be noted, however, that the radiographic error model presented here includes parameters for positional misalignments of the detector. Positional misalignments in X and Y can be modelled by adapting the principal point u_o and v_o, whereas a misalignment in Z is modelled by adapting SDD.

A detector can be misaligned by more than one rotation angle simultaneously. Various established conventions may be used for rotating three-dimensional coordinates [7]; these conventions differ by the axes about which the rotations are performed and the sequence in which the axes are rotated. In general, the application of different conventions will not generate equivalent final three-dimensional rotations. The convention used here is chosen to agree with the convention used to simulate a rotation of the detector in the analytical (ray-tracing) simulation software Scorpius XLab®. More information on Scorpius XLab® can be found in the literature