Estimation of the Source-Detector Alignment of Cone-Beam X-ray Systems using Collimator Edge Tracking

Christoph Luckner, Thomas Mertelmeier, Andreas Maier, and Ludwig Ritschl

***Abstract*—Upcoming applications for clinically well-established digital X-ray systems, like workflow automation, high-quality free exposures with mobile detectors, free tomosynthesis or weight-bearing full-body acquisitions with dynamic wireless detectors require a precise and reproducible method to determine the source-detector alignment. This alignment is usually obtained once in a calibration step through the imaging of a phantom with known marker geometry or via online-calibration methods, which are currently still subject to research. The former approach usually suffers from a degrading image quality over time and the latter complicates the clinical workflow due to the cumbersome and prone-to-error positioning of the patient along the additional hardware. The proposed three-step method, in contrast, only uses the existing collimator of the X-ray system and its projection and does not require any additional hardware. We assume that the extrinsic projection parameters and the orientation of the source are already known considerably well by the system, while the intrinsic projection parameters still have to be estimated individually for each scan. For evaluation, we compared the result of the proposed method to the parameters obtained through the imaging of a calibration phantom. It could be shown that the proposed method is able to achieve a high accuracy for the estimation of the intrinsic projection parameters, i. e. focal length and principal point, with a mean relative error lower than 0.5 %.**

I. Introduction

In the clinical field, X-ray imaging and X-ray computed tomography (CT) is widely used for visualizing the inside of the human body. With upcoming applications for those clinically well-established digital X-ray systems, like workflow automation, high-quality free exposures with mobile detectors, or free tomosynthesis acquisitions with dynamic wireless detectors a precise and reproducible method to determine the source-detector alignment is required. Moreover, novel X-ray systems with independently movable source and detector like the Multitom Rax (Siemens Healthcare) which can be seen in Figure 1a might benefit from the presented method. Usually, the source-detector alignment is determined once in a calibration step through the imaging of a phantom with known marker geometry [1], [2]. However, one drawback of such a one-time calibration is that the behavior of the system changes over time, which leads to a degrading image quality. There also exist several online calibration methods which rely on either phantoms or fiducial markers [3], [4], [5], other calibration objects [6] which have to be visible in the acquired image or are purely image-based [7]. However, the positioning of the patient along the additional hardware is a cumbersome and prone-to-error process and complicates the clinical workflow.

The proposed method, in contrast, uses only the already existing collimator of the X-ray system and its projection to estimate the source-detector alignment and does not require any additional hardware.

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II. Materials and Methods

Any arbitrary 3-D point p^{3D} in a world coordinate system can be mapped onto the 2-D detector plane of an X-ray imaging system using a projection matrix $\mathbf{P}$

$$p^{\mathrm{2D}} = \mathbf{P} \cdot p^{\mathrm{3D}} = \mathbf{K}\,[\mathbf{R}\ \mathbf{t}] \cdot p^{\mathrm{3D}}, \tag{1}$$

where $\mathbf{R}$ and $\mathbf{t}$ denote the extrinsic parameters of the imaging system, i. e. rotation and translation form the world coordinate system to the detector coordinate system, and $\mathbf{K}$ the intrinsic parameters, i. e. focal length and the principal point, as illustrated in Figure 1b.

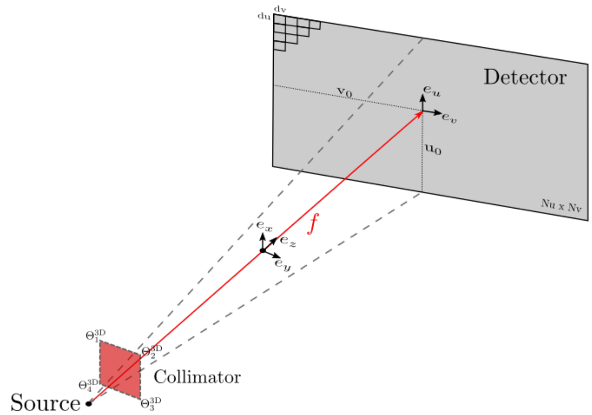
We assume, that the position of the imaging system relative to a point in the exam room is known. Hence, rotation and translation are considered to be known, whereas the intrinsic parameters still have to be estimated.

A. Algorithm

In the following a three-step algorithm to estimate the source-detector alignment using only the collimator of the imaging system will be presented.

1) Corner detection:
   In the first step the corners of the collimator c^{2D} in the X-ray image I have to be detected. This can either be done by corner detection algorithms like the Harris corner detector [8] or by an intersection of the collimator edges.
   Due to the fact that not it might happen, that not all corners of the collimator might be visible in the image, we decided to use the second approach. Therefore, we computed the Hough-transform $H(I)$ of the input image I to detect the outline of the collimator, indicated as colored lines in Figure 2. Afterwards, the intersection

(a) Twin-robotic X-ray system Multitom Rax (Siemens Healthcare) with two independently movable ceiling-mounted robotic arms for source and detector.



(b) Schematic illustration of acquisition geometry. The focal length f is indicated as red arrow, the point where the principal ray hits the detector perpendicularly is called principal point (u_0, v_0). The collimator of the X-ray system is indicated as semi-transparent red square.

Fig. 1. The used imaging system Multitom Rax (Siemens Healthcare) and a schematic illustration of the acquisition geometry.

of all lines with each other was computed, which results in the four corners of the collimator $c^{2\text{D}}$.

2) Initial estimate:
The detected corners $c^{2\text{D}}$ were then used as input to a least square approximation in order to compute an initial estimate for the focal length $\dot{f}$ and the principal point $(\dot{u}_0, \dot{v}_0)$

$$\dot{f}, \dot{u}_0, \dot{v}_0 = \underset{f,u_0,v_0}{\arg\min} \sum_{i=1}^{n} |\mathbf{P}(f, u_0, v_0) \cdot \Theta_i^{3\text{D}} - c_i^{2\text{D}}|_2^2, \tag{2}$$

where $\Theta_i^{3\text{D}}$ denotes the i-th corner of the collimator in the world coordinate system.
This step was performed using a grid search approach, which exhaustively considered all possible parameter combinations.

3) Refinement step:
Since the corner detection itself might be prone to errors, e. g. due to not clearly visible collimator edges, we propose to use a refinement step which does not rely on the detected corners but uses the gradient information in the image. First, the gradient magnitude image G of the input image I has to be computed

$$G(u, v) = \sqrt{g_u^2 + g_v^2}, \tag{3}$$

with

$$\nabla I = \begin{bmatrix} g_u \\ g_v \end{bmatrix} = \begin{bmatrix} \frac{\partial I}{\partial u} \\ \frac{\partial I}{\partial v} \end{bmatrix}. \tag{4}$$

Then, based on the set of initial parameters $(\dot{f}, \dot{u}_0, \dot{v}_0)$ from the previous step, we project two adjacent collimator points $\Theta_i^{3\text{D}}$ onto the detector and evaluate the line integral E between those two points utilizing the fact that there are only vertical and horizontal edges

$$E(p_i^{2\text{D}}, p_{i+1}^{2\text{D}}) = \int_{p_i}^{p_{i+1}} G(u, v)\ \mathrm{d}u\ \mathrm{d}v \tag{5}$$

and

$$p_i^{2\text{D}} = \mathbf{P}(f, u_0, v_0) \cdot \Theta_i^{3\text{D}}. \tag{6}$$

Since the gradient magnitude has its maximum at the exact location of the edge (see Figure 3), we aim to maximize the sum over all line integrals, in order to obtain a refined estimate $(\ddot{f}, \ddot{u}_0, \ddot{v}_0)$ of the intrinsic parameters

$$\ddot{f}, \ddot{u}_0, \ddot{v}_0 = \underset{f,u_0,v_0}{\arg\max} \sum_{i=1}^{n} E(p_i^{2\text{D}}, p_{i+1}^{2\text{D}}). \tag{7}$$

B. Experiment: Proof of Concept

In order to evaluate the proposed method, we conducted the following experiment. We acquired 3-D scans with the twin robotic X-ray system Multitom Rax (Siemens Healthcare) consisting of 152 projections along a circular trajectory.

Each scan was performed twice: first with calibration phantom (PDS-2 phantom [1]) which was later used to determine the ground truth and a second scan without phantom. We used a Matlab implementation of the proposed method to estimate the intrinsic parameters and evaluated the relative error $\varepsilon(p)$ for each intrinsic parameter $p \in \{f, u_0, v_0\}$ separately

$$\varepsilon(p) = \frac{p_{\text{GT}} - \ddot{p}}{p_{\text{GT}}}. \tag{8}$$

For an additional visual inspection, we performed a forward projection of the collimator edges with both the ground truth projection matrix and the reassembled projection matrix with the estimated intrinsic parameters and compared the results to each other.

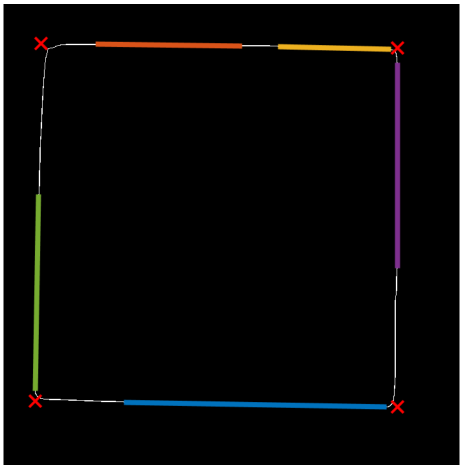


Fig. 2. Hough transformed image $H(I)$ with detected edges (colored lines) and the computed corners (red crosses).

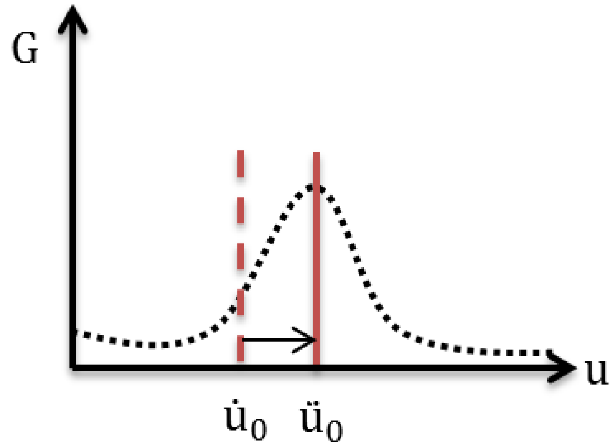


Fig. 3. Schematic drawing of the refinement step. The initial estimate $\dot{u}_0$ is moved towards the maximum of the gradient magnitude G, which yields that $\ddot{u}_0$ finally ends up at the exact location of the edge.

III. Results

Figure 4 shows the relative errors ε for each intrinsic parameter. An overall high accuracy with a mean relative error of $\bar{\varepsilon}(f) = 0.29\,\%$, $\bar{\varepsilon}(u_0) = 0.06\,\%$, and $\bar{\varepsilon}(v_0) = 0.43\,\%$ could be achieved. The computation time for each projection was about 4 seconds in a non-optimized CPU implementation. Furthermore, an exemplary result after each step as described in Section II-A can be seen in Figure 5.

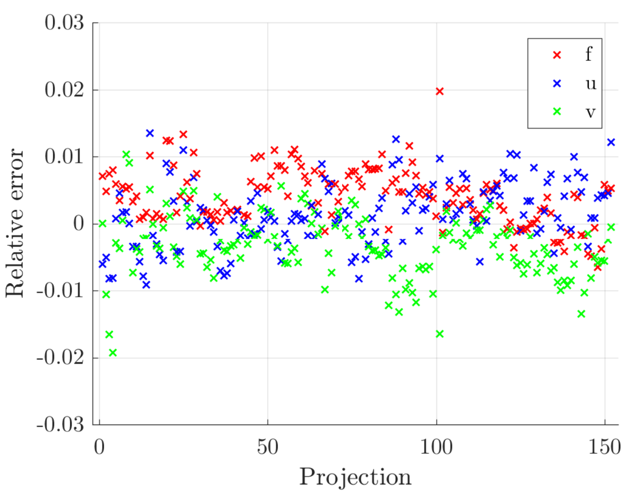


Fig. 4. Plot of the relativ error for each intrinsic variable, $\varepsilon(f)$ in red, $\varepsilon(u_0)$ in blue, and $\varepsilon(v_0)$ in green

IV. Conclusion

The presented algorithm is capable of estimating the source-detector alignment of cone-beam X-ray systems, utilizing only already existing information of the X-ray system. The proposed method might open up the possibility of further workflow automation and image quality improvement in well-established digital X-ray systems. In terms of online calibration, this method enables free tomosynthesis acquisitions in case of an exactly known detector position and orientation. Moreover, especially purely line-based trajectories where source and detector move simultaneously in parallel planes [9] can benefit from such a method since the orientation of the source as well as the extrinsic parameters of the system remain nearly constant during the entire scan. Additionally, since the method is purely image-based we do not introduce any complications in clinical workflow since no additional hardware, such as calibration phantoms or markers are required. As the initial estimation of the source-detector alignment relies on the detected corners of the collimator, a more sophisticated algorithm which is already used for auto-cropping of X-ray images [10] might lead to further improvements in the estimates.

As a topic for future research, instead of using the grid search approach in the second stage of the algorithm, a solution using, for instance, an SVD approach, could speed up the algorithm.

Disclaimer

The presented method is commercially not available, its availability cannot be guaranteed. The Siemens Healthineers Multitom Rax is not available in all countries, its availability cannot be guaranteed.

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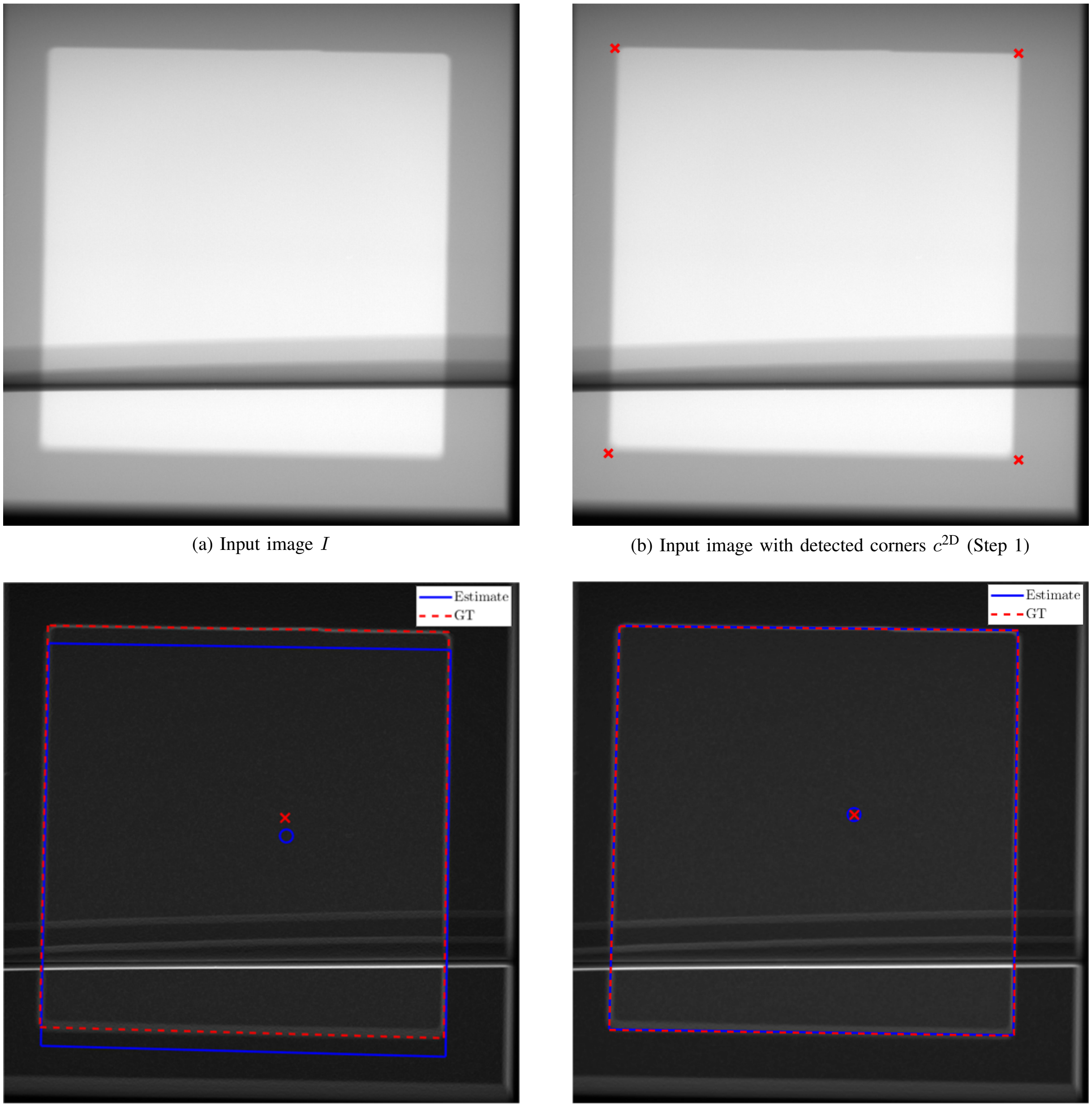

(a) Input image I

(b) Input image with detected corners c^{2D} (Step 1)

(c) Initial estimate (Step 2)

(d) Refined estimate (Step 3)

Fig. 5. Input and output images. The forward projection of the collimator using the ground truth projection matrix is indicated as red dashed line, the estimated one as solid blue line.

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