Estimation of Guidewire Inclination Angle for 3D Reconstruction

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Abstract—**Reconstruction of the 3D position of the guidewire is an important step toward automated navigation systems supporting minimally invasive endovascular interventions. We present a method to estimate the projected guidewire thickness for the monoplane X-ray configuration and we propose a novel method to recover the local angle between the guidewire and the X-ray path and to recover the inclination angle of the guidewire to the imaging plane, up to an orientation. We experimentally show the feasibility of the proposed approach by recovering both the local and the inclination angle of the guidewire; the average error for the recovered inclination angle is 8.6±8.2°.**

Keywords—**X-ray imaging, guidewire, inclination angle, thickness estimation.**

I. INTRODUCTION

During a minimally invasive endovascular intervention a guidewire is introduced into the vasculature and must be navigated to a point of interest, usually an aneurysm, a stenosis or an arteriovenous malformation. Accurate navigation through the vessels is a prerequisite for a successful intervention. Fluoroscopic X-ray imaging is currently used to aid in navigation, however, physicians still interpret obtained 2D fluoroscopic images themselves. Robust recovery of the 3D position from a single fluoroscopic image would represent a significant navigational improvement.

Either monoplane or biplane configuration is used for navigation. Position reconstruction for the biplane configuration is straightforward [1], but for the monoplane configuration there is an inherent ambiguity that must be overcome by using some additional knowledge. In previous works by Van Walsum et al. [2] and Petković et al. [3] an optimization method relying on known position of the blood vessels is used. Esthappan et al. [4] uses 2D/3D registration to align the model to the projection of the pre-shaped guidewire tip. Abovementioned methods only extract the guidewire position and local orientation from the 2D fluoroscopic images.

In this paper, we present a technique to estimate the projected guidewire thickness and we present a novel extension to recover the local angle between the guidewire and the X-ray path, up to an orientation, and to recover the guidewire inclination angle toward the image plane, up to an orientation, using a single fluoroscopic image. The inclination angle is significant for the reconstruction of the 3D position of the guidewire as it, together with the local 2D direction vector in the image plane, defines the local 3D direction vector of the guidewire.

The paper is structured as follows: in Section II a simple X-ray imaging model is presented; in Section III we show how to estimate the thickness and present the method to estimate the inclination angle; results are presented in Section IV; we conclude in Section V.

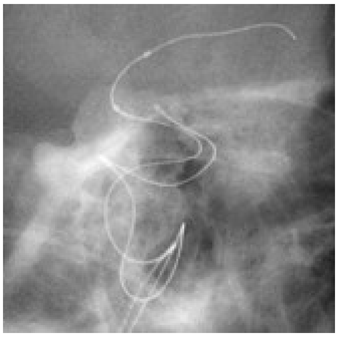

Fig. 1 A fluoroscopic image showing guidewires during a neurointervention.

II. IMAGING MODEL

If a homogeneous and isotropic object is placed between the X-ray source and the detector then the observed intensity can be approximated by [5,6]

$$I = I_0 \exp\left(-\sum_i \mu_i d_i\right) = I_0 \prod_i \gamma_i \,, \tag{1}$$

where μ_i [cm^{-1}] is the linear attenuation coefficient and γ_i is the multiplicative attenuation factor of an i-th object on the X-ray path. The multiplicative model (1) is valid for a narrow energy band and disregards the beam diffusion. Any factor of the quantity $\prod_i \gamma_i$ may model the guidewire. Let γ_{gw} denote the multiplicative term of the guidewire. Then for pixels where the guidewire is absent we observe the intensity

$$I_1 = I_0 \prod_i \gamma_i \ , \tag{2}$$

and for pixels where the guidewire is present we observe the intensity

$$I_2 = I_0 \exp\left(-\sum_i \mu_i d_i\right) \exp\left(-\mu_{\mathrm{gw}} d_{\mathrm{gw}}\right) = I_1 \gamma_{\mathrm{gw}} \ . \tag{3}$$

The imaging model is multiplicative in intensity and is summarized by Equations (2) and (3). The multiplicative nature causes the contrast variation along the length of the guidewire; from this contrast variation the projected thickness and the inclination angle can be recovered using a single fluoroscopic image.

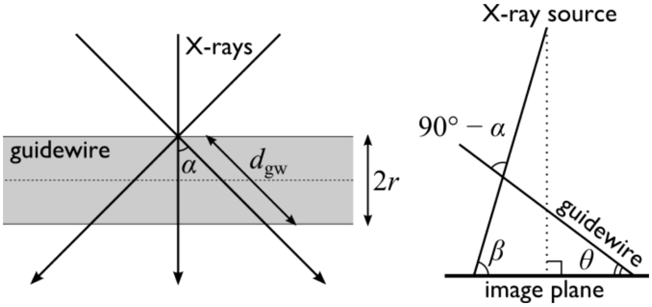

Fig. 2 Geometry of an intersection between the guidewire and an X-ray. Note that the solution is not unique, there are two rays that yield the same value of projected thickness d_{gw} .

III. THICKNESS AND ANGLE RECOVERY

Let's assume the values (2) and (3) are known. Using division the multiplicative term γ_{gw} can be recovered,

$$\frac{I_2}{I_1} = \frac{I_1 \gamma_{\mathrm{gw}}}{I_1} = \gamma_{\mathrm{gw}} \ , \tag{4}$$

and an estimate of the projected guidewire thickness d_{gw} can be obtained as

$$d_{\mathrm{gw}} = -\frac{1}{\mu_{\mathrm{gw}}} \log \gamma_{\mathrm{gw}} \ . \tag{5}$$

Note that d_{gw} given by (5) is the length of the path X-rays travel to pass through the guidewire and is not equal to the guidewire diameter $2r$ (Fig. 2). Also, to recover d_{gw} in real-world units the value of μ_{gw} [cm^{-1}] must be known, thus used imaging system must be calibrated and guidewire parameters must be known. However, if we are interested in recovering only local angles between the guidewire and X-rays then the value of μ_{gw} [cm^{-1}] is not needed.

Consider the geometry shown in Fig. 2. To recover the local angle α between the guidewire and an X-ray we use

$$2r = d_{\mathrm{gw}} \cos(\alpha) \ , \tag{6}$$

so the diameter $2r$ must be known. However, by using (5) we obtain

$$\cos(\alpha) = \frac{2r}{d_{\mathrm{gw}}} = \frac{\log \gamma_{\mathrm{gw,min}}}{\log \gamma_{\mathrm{gw}}} \ , \tag{7}$$

and the term μ_{gw} disappears, therefore the angle α can be recovered from the un-calibrated data. The value of $\gamma_{\mathrm{gw,min}}$ that is tied to the diameter $2r$ can be estimated using the same data.

Once the local angle α is known the inclination angle θ can be obtained as

$$\theta = 90^{\circ} + \alpha - \beta \ , \tag{8}$$

but only up to an orientation, as the guidewire can have either falling or raising slope (Fig. 2). The angle β is the angle between X-ray and the image plane and is determined by the C-arm geometry.

The value I_2 is measured directly from the input X-ray image from pixels that are identified as the guidewire by some segmentation procedure, e.g. as described in [7,8,9, 10]. Then, using the values of adjacent background pixels the value I_1 must be estimated as it cannot be measured directly.

Therefore, the proposed method to recover the inclination angle of the guidewire from a single X-ray image is comprised of the following steps:

1. segmentation of the guidewire in the X-ray image and measurement of intensity values I_2 ,
2. estimation of the background values I_1 using intensity values adjacent to pixels identified in step 1,
3. computing the ratio (4) and estimating the value of $\gamma_{\mathrm{gw,min}}$, and
4. estimation of angles α and θ using Eqs. (7) and (8).

For the step one of the implementation we use our previous work, the adaptive vesselness measure [9], which yields both the segmentation and the local guidewire direction in 2D. Combining this information with the estimated θ defines the local guidewire direction vector in 3D. Background I_1 is estimated as an arithmetic mean of adjacent pixels on both sides of the segmented guidewire.

IV. RESULTS AND DISCUSSION

To determine if the proposed estimation scheme for the inclination angle can be used a trial phantom image was acquired. Acquisition was done using Philips Allura X-ray imaging system with pixel size 0.217 mm and 1016×1016 image resolution. The guidewire was placed so it forms a large circle that is located mostly in a plane orthogonal to the imaging plane, starting with an inclination angle of about 40° sloping downward toward the image plane (Fig. 3). For this placement we expect the projected thickness to vary along the guidewire length thus indicating the local guidewire orientation relative to X-ray paths, and to cover whole range of inclination angles of interest.

The segmented guidewire in 2D is shown in Fig. 4 (computed as described in [9]). Detail of the segmentation is shown in Fig. 5: the value I_2 is measured directly from the input X-ray image at the guidewire centerline (yellow); the value I_1 is estimated as arithmetic mean of adjacent pixels on both sides of the guidewire (red). Note that if an X-ray image sequence is available then the Kalman filter approach of [10] can be used for improved temporal estimation of the background intensity I_1.

The projected thickness estimate computed directly using I_1 and I_2 must be smoothed as the initial estimate based on single pixel values is too noisy for practical use (Fig. 6, thin line). This effect is both due to present noise and due to suboptimal placement of the guidewire centerline (pixel instead of sub-pixel precision). However, applying a smoothing spline with parameter $p = 0.01$ significantly improves the result (Fig. 6, thick line). Note the thickness estimate is noisy but is varying along the length of the guidewire as expected; there are two sharp increases in the estimated thickness corresponding to the raising and the falling slopes of the circle the guidewire forms.

The value of $\gamma_{\text{gw,min}}$ must be estimated from the projected thickness estimate. For this experiment we have manually estimated $\gamma_{\text{gw,min}} = 0.9$ using the value at the central position between two peaks; this position corresponds to the segment parallel to the imaging plane. Note that due to noise and disregarded partial volume and beam diffusion effects the value of 0.9 is not the minimum value so cos of Eq. (7) is not well-defined for all data points.

For the practical use of the proposed technique the estimation procedure for $\gamma_{\text{gw,min}}$ must be automated: if the position of the blood vessels is known then it may be estimated using the known blood vessel angle and the assumption the guidewire has the same direction as blood vessels.

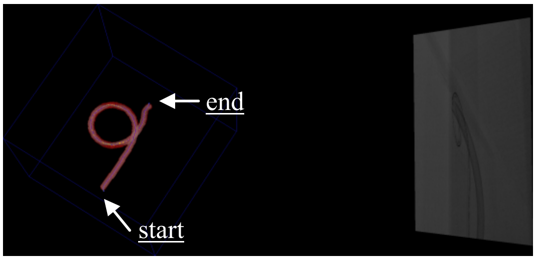

Fig. 3 Imaging geometry (3DRA reconstruction, imaging plane).

Fig. 4 Segmented guidewire (adaptive vesselness map).

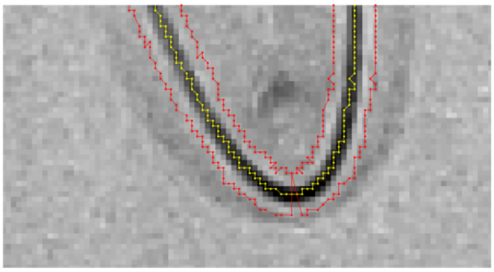

Fig. 5 Detail showing segmented guidewire centerline and adjacent pixels on both sides of the guidewire (transition area of about 3 px excluded). Note the contrast variation along the length of the guidewire and the peak at the point of maximum curvature.

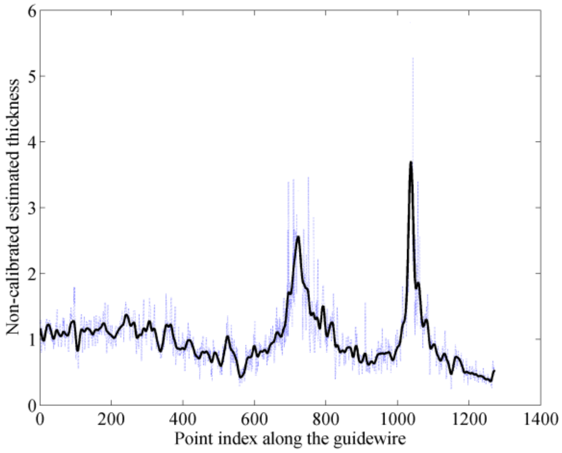

Fig. 6 Smoothed thickness estimate (initial estimate is shown as thin blue line).

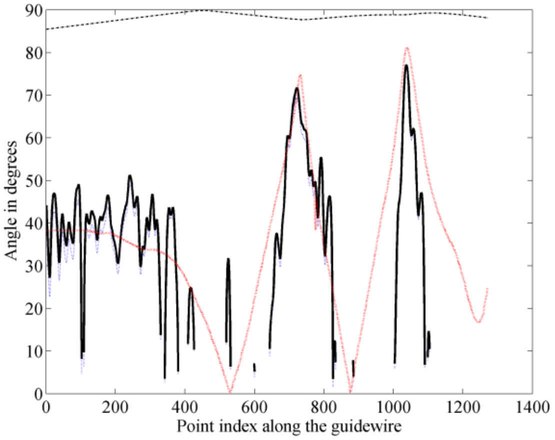

Fig. 7 Estimated angles: local angle α between the guidewire and an X-ray is represented by a thin blue line, X-ray angle β is represented by a dotted black line, and inclination angle θ is represented by a thick black line (estimation) and thin red line (true value).

Once $\gamma_{\mathrm{gw,min}}$ is known the inclination angle can be estimated. Estimated angles are shown in Fig. 7, starting at about 40°, then dropping toward 0° as the loop starts etc. The true value of the angle recovered using 3DRA reconstruction is shown in red. A total of 1272 pixels is segmented as the guidewire; inclination angle estimation succeeded for 678 pixels and failed for 594 pixels due to Eq. (7) being larger than one. For data points where estimation succeeded the average inclination angle error is 8.6±8.2°, but the maximum error is 52.5°. The average error is a promising first result for application in 3D reconstruction, especially as further improvements are expected. However, further post-processing is needed to reduce the maximal error.

V. CONCLUSIONS

We have presented a novel estimation technique to recover the local angle between the guidewire and the X-ray path, up to an orientation, and to recover the guidewire inclination angle toward the image plane, up to an orientation, using a single fluoroscopic image. Using this information a guidewire direction in 3D can be recovered and used as an aid in the 3D position reconstruction thus further improving upon existing 3D reconstruction techniques described in the literature.

Future work should focus on improving the estimation of I_1 and I_2 using sub-pixel precision and the Poisson noise formalism. Furthermore, partial volume effects and beam diffusion must be included in the imaging model.

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CONFLICT OF INTEREST

The authors declare that they have no conflict of interest.

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