X-IVUS:
Integrated x-ray and IVUS system for the Cathlab

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ABSTRACT

Percutaneous Transluminal Coronary Angioplasty is currently the preferred method for coronary artery disease treatment. Angiograms depict residual lumen, but lack information about plaque characteristics and exact geometry. During instrument positioning, intracoronary characterization at the current instrument location is desirable. By pulling back an intravascular ultrasound (IVUS) probe through a stenosis, cross-sections of the artery are acquired. These images can provide the desired characterization if they are properly registered to diagnostic angiograms or interventional fluoroscopies. The method we propose acquires fluoroscopy frames at the beginning, end, and optionally during a constant speed pullback. The IVUS probe is localized and registered to previously acquired angiograms using a compensation algorithm for heartbeat and respiration. Then, for each heart phase, the pullback path is interpolated and the corresponding IVUS frames are positioned. During the intervention the instrument is localized and registered onto the pullback path. Thus, each IVUS frame can be registered with a position on an angiogram or to an instrument location and during subsequent steps of the intervention the appropriate IVUS frames can be displayed as if an IVUS probe were present at the instrument position. The method was tested using a phantom featuring respiratory and contraction movement and an automatic pullback with constant speed . The IVUS acquisition was replaced by fibre optics and the phantom was imaged in angiographic and fluoroscopic modes. The study showed that for the phantom case it is indeed possible to register the IVUS cross-section to the interventional instrument positions to an accuracy of less than 2mm.

Keywords: IVUS to x-ray registration, image registration, respiration compensation, PTCA.

1. INTRODUCTION

Percutaneous Transluminal Coronary Angioplasty (PTCA) is currently the preferred method for treating coronary artery disease. The procedure is mostly performed in monoplane Cathlab systems, which provide projection x-ray images. For diagnosis, a radio-opaque contrast dye is injected into the coronary arteries and sequences of angiographic images with different geometry settings are acquired. A stenosis is identified as a narrowing of the visible lumen. One of these angiographic frames is displayed on a monitor during intervention and serves as roadmap to subsequently navigate the instruments to the lesion. It depends on the physician's ability to fuse the information provided in the roadmap and the live fluoroscopic images on another monitor to position the instrument during treatment (Figure 1). Inaccurate positioning of balloons or stents is related to high re-stenosis ratios.[1]

The current workflow during Cathlab interventions relies heavily on angiography for diagnosis and localization of stenoses. Angiograms provide an excellent overview on the whole artery tree, but do not provide local information about the type of plaque or its exact geometry. In contrast, Intravascular Ultrasound (IVUS) is a local imaging modality that gives information about the plaque quality as well as on the geometry of the stenosis. By pulling back an IVUS probe

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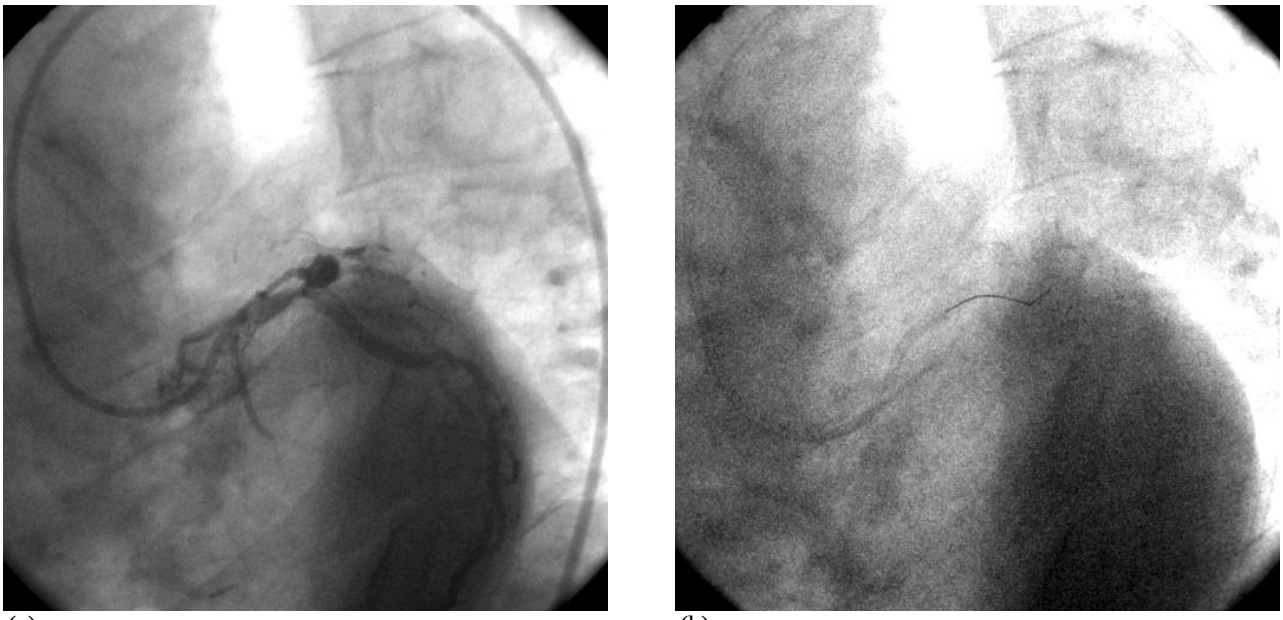
from the distal to the proximal end of a lesion, cross-sectional views from within the artery along the pullback path are obtained. Nevertheless, IVUS does not provide information about the vessel topology or geometry, which prevents an accurate reconstruction of the vessel based only on IVUS.



(a) (b)

Figure 1. Cathlab monitors: on the left monitor, the static angiographic roadmap displaying the contrast-filled vessels (a) and on the right monitor the live fluoroscopic frame where the guidewire tip is visible (b).

By fusing projection x-ray and IVUS images, the ability to provide accurate geometric information about the local (IVUS) and global (x-ray) shape of the vessel is combined. The resulting information can be used both for diagnosis[2,3,4] and for interventions to allow a more precise treatment (e.g. stent placement or brachytherapy). The aim of the method proposed in this paper is twofold: First, to improve diagnosis and treatment planning by fusion of angiography and IVUS derived information and second, to aid the positioning of the interventional instrument by displaying at all times the IVUS frame that corresponds to the current instrument position. The method is based on the characterisation of each acquired x-ray image by a heart and a respiration phase. These two variables define the heart position and state in each image.[6,7] By registering images with the same heart and respiration phases, it is possible to relate structures in them, e.g. the contrasted coronary vessels in an angiogram with the guidewire tip visible in an interventional frame. Using our registration method and a respiration compensation method[6,7,8] it is possible to register the IVUS probe position in a pullback sequence with previously acquired angiograms to provide a diagnostic tool, and to register live interventional images with previously acquired pullback frames, therefore providing the IVUS image that was acquired at the current instrument position.

This paper presents a novel integration of IVUS into the x-ray Cathlab interventional workflow. Whereas previous work[2,3,4,5,10] proposes a 3D reconstruction of the whole artery imaged with biplane angiography, and thus provides a 3D diagnostic tool that accurately quantifies the characteristics of the plaque, in this work we present a more practical approach that provides the physician with a diagnostic and an interventional tool. To the author's knowledge, no system has been proposed to date that employs previously acquired IVUS information during the intervention.

The document is structured as follows: in Section 2 the required registration method is explained, in Section 3 our experiments with a heart phantom are introduced with results presented in Section 4. Conclusions regarding our proposed methods and experimental results are given in Section 5.

2. METHODOLOGY

In the current clinical workflow, one or several angiographic sequences with different system geometry settings are acquired to diagnose a coronary lesion. We propose to allow the patient to breathe freely during the acquisition of contrasted coronary angiograms, is simpler for the patient and allows us to calculate a respiration phase for each acquired angiogram and to estimate the motion of coronaries due to respiration. The physician chooses an imaging geometry in which the stenosis is clearly visible and navigates a guidewire under fluoroscopic survey to the lesion. An IVUS probe is then positioned at the distal end of the lesion, the IVUS catheter is stretched a little before starting the pullback,[4,5] and a constant speed pullback begins while IVUS images and the ECG are captured by a suitable system. To provide the method presented here with required information on the position of the IVUS probe, fluoroscopic frames are acquired at the beginning, end, and optionally during its constant speed pullback of the IVUS probe.

All angiographic and fluoroscopic frames are characterised by the heart contraction and the respiration state at the time instant when they were acquired.[6,7] The heart phase is calculated from synchronously acquired ECG, while the respiration phase is calculated from the image content as a similarity measure[9] between every image in the sequence and an angiographic reference selected to show end inspiration. Each image can thus be characterised by these two variables and be represented as a point in a two-dimensional *phase-space*.[6,7] Nearby points represent images with similar contents, i.e., images that present the heart at the same position and contraction state. Because of the low breathing frequency, it is common that the phase-space is not densely sampled along the respiration phase axis. Respiration compensation algorithms have been reported previously.[7,8] By characterising each frame with a respiration and a heart phase, and by employing respiration compensation algorithms, we are able to register any two frames (diagnostic or interventional) in terms of heart contraction state and position. We use this method to register diagnostic angiograms with pullback fluoroscopy frames for diagnostic purposes (section 2.1) and interventional and pullback fluoroscopy frames for accurate positioning (Sec. 2.2).

2.1 X-ray and IVUS fusion for diagnosis

The objective of fusing pre-interventional projection x-ray angiograms with IVUS is to provide the physician with a diagnostic tool that relates to each point along a diseased vessel an IVUS image (Figure 2). The main difference with previous approaches [2,3,4,5] is that we use single projection x-ray images to present the coronary geometry and, therefore, avoid the need of a biplane system, which is not of common use during PTCA interventions.

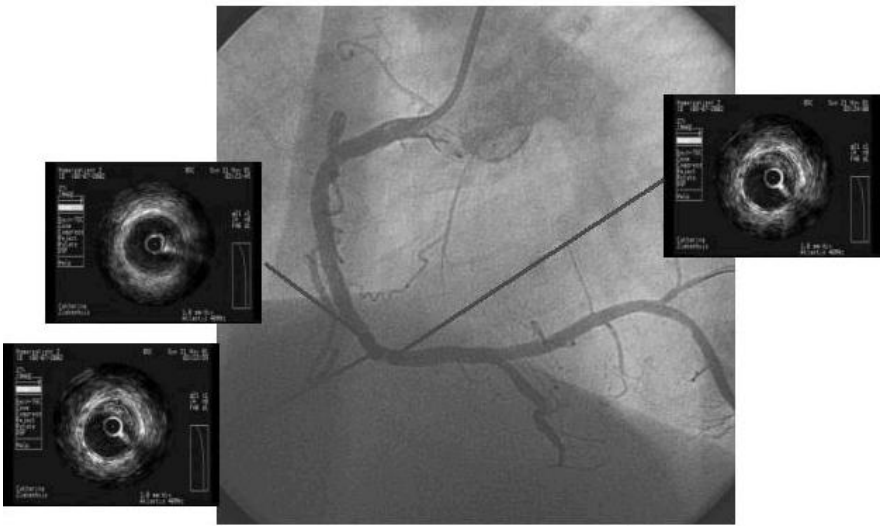


Figure 2. Exemplary angiography registered with an IVUS pullback. The pullback path is identified, and IVUS frames are assigned to each point along this path.

Using the phase-space and a respiration compensation algorithm, we register the diagnostic and pullback frames. As a first step, an angiographic sequence with the appropriate geometry settings is selected. The heart phase is quantised in *N* possible values, and for each heart phase δ_n, an angiogram $A(\delta_n,\beta_n)$ is selected. Criterion for this automated selection is a high amount of contrast agent visible in these frames to provide suitable geometry information to the user[14]. Next, the fluoroscopy x-ray frames from the pullback are registered with the reference angiogram frames with the same respective heart phase. Here, the motion of coronaries due to respiration is compensated as described below. This allows to reconstruct a pullback path for each heart phase. Each fluoroscopy pullback frame $P(\delta_j,\beta_j,x_j,y_j)$ is characterised by its heart and respiration phase, δ_j and β_j respectively, and the IVUS probe position in the frame (x_j,y_j), which is detected by radio-opaque markers on the probe. Each IVUS frame is characterised by a time stamp that relates it to the synchronously acquired fluoroscopy sequence.

A displacement vector from the reference angiogram with same contraction phase δ_n but different respiration phase β_n is computed for the fluoroscopy frame acquired with a respiration phase β_j. The angiogram phase shows the coronaries in the same contraction phase but shifted due to respiration motion.

$$\vec{m}_j^n = (m_{x,j}^n, m_{y,j}^n) = R(A(\delta_n, \beta_n), P(\delta_n, \beta_j, x_j, y_j)) \quad (1)$$

where

- R is a respiration compensation function [7,8],
- $\vec{m}_j^n = (m_{x,j}^n, m_{y,j}^n)$ is the displacement vector that relates the position of the heart within the reference angiogram and within the given pullback fluoroscopy frame,
- $A(\delta_n, \beta_n)$ is the angiogram selected for the corresponding heart phase δ_n, and
- $P(\delta_n, \beta_n, x_j, y_j)$ is a pullback frame with heart phase δ_n.

Once the translation vector is known for each pullback frame, the respiration-compensated position of the IVUS probe $(\hat{x}_j, \hat{y}_j)$ is calculated as

$$(\hat{x}_j, \hat{y}_j) = \vec{m}_j^n + (x_i, y_i), \quad (2)$$

which serves as one vertex in the creation of the pullback path. The respiration-compensated position of the probe registers it with its corresponding position in the reference angiography. If only the beginning and the end of the pullback are available, the pullback path $\Pi(\delta_j, \beta_j)$ is reconstructed from these two points and the vessel segment visible in the angiogram by calculating its centreline [11,12] and taking into account the constant pullback speed. If the pullback was constantly imaged with fluoroscopy, then the pullback path can be reconstructed by interpolation between the probe positions and the IVUS images are located by means of the known pullback speed.

2.2 X-ray and IVUS fusion for accurate positioning

In spite of the extensive literature on 3D vessel reconstruction fusing IVUS with angiography, no integrated system has yet been developed (to the authors' knowledge) that can be applied during the intervention itself. With the proposed algorithm, we can register each interventional fluoroscopy frame displaying the current instrument position at a certain heart and respiration phase with the previously reconstructed pullback path for the same heart phase, compensating for the difference in respiration phase. Thus we are able to relate said position to the IVUS frame acquired at that location for the same heart phase. This registration algorithm is executed in real-time for each acquired fluoroscopy frame.

For each incoming live fluoroscopic framee its respiration and heart phase, and the instrument location are calculated in real-time $F(\delta_n, \beta_i, x_i, y_i)$.[13,15] A translation vector $\vec{m}_i$ is calculated to register the interventional frame to the corresponding pullback path $\Pi(\delta_n, \beta_n)$ acquired with the same contraction phase using respiration compensation:

$$\vec{m}_i = R(\Pi(\delta_n, \beta_n), F(\delta_n, \beta_i, x_i, y_i)) \quad (3)$$

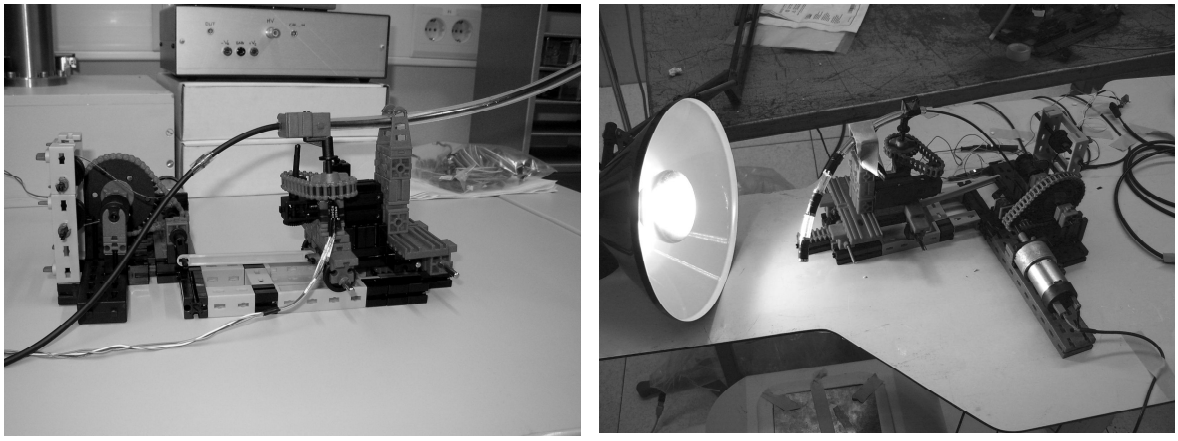
Given $\vec{m}_i$, again the respiration-compensated position of the instrument $(\hat{x}_i, \hat{y}_i)$ can be calculated as

$$(\hat{x}_i, \hat{y}_i) = \vec{m}_i + (x_i, y_i) \quad (4)$$

$(\hat{x}_i, \hat{y}_i)$ is a position within $\Pi(\delta_n, \beta_n)$ and therefore, linked to an IVUS frame acquired at this position. By registering the current instrument position with previously acquired IVUS images, a support for the accurate positioning of the instrument can be given.

3. EXPERIMENTS

To test the performance of our method in assigning a proper IVUS image to each point along a vessel and in registering the current instrument position with a previously acquired IVUS image, we used a phantom built with Fischer Technik®[*] featuring independent respiratory and contraction movements and constant speed automatic pullback (Figure 3a). The IVUS pullback acquisition was replaced by a fiber optic with a radio-opaque marker (Figure 3b) and a bright illumination source. A coronary vessel was simulated with a transparent plastic tube with an internal diameter of 5mm and with alternate opaque sections of known length, which were marked with partially opaque metal bands (Figure 4). Note that this phantom provides radio-opaque markers visible on x-ray acquisitions to locate the probe and to define position and runlength along the vessel. The runlength of the fiber optic is further defined by the succession of dark and bright bands such that a ground truth for the registration is available.



(a) (b)

Figure 3. Heart phantom featuring cyclic respiration- and heart contraction-mimicking movement (a). The phantom allows a constant-speed pullback of a fiber optic. Experimental setup to simulate IVUS acquisition by a fiber optic gathering light from an illumination source (b).

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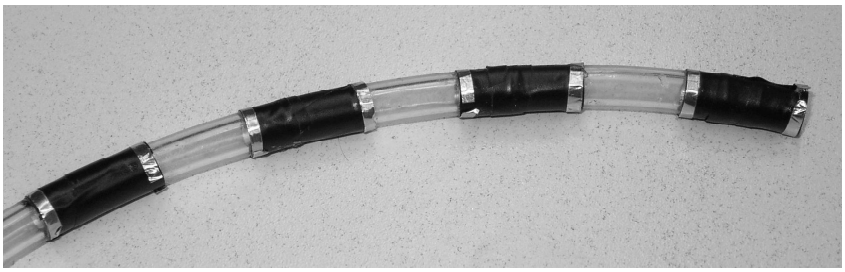
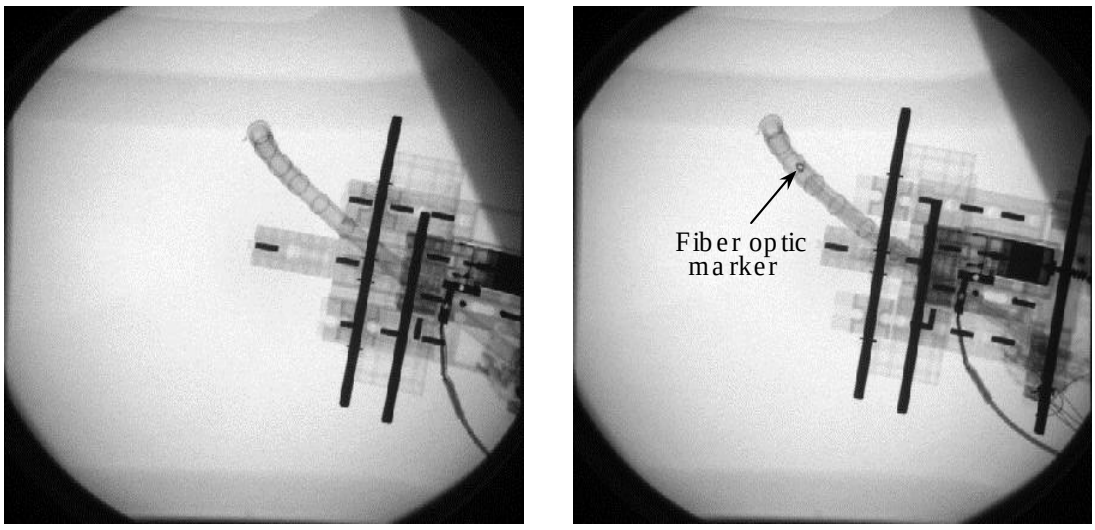


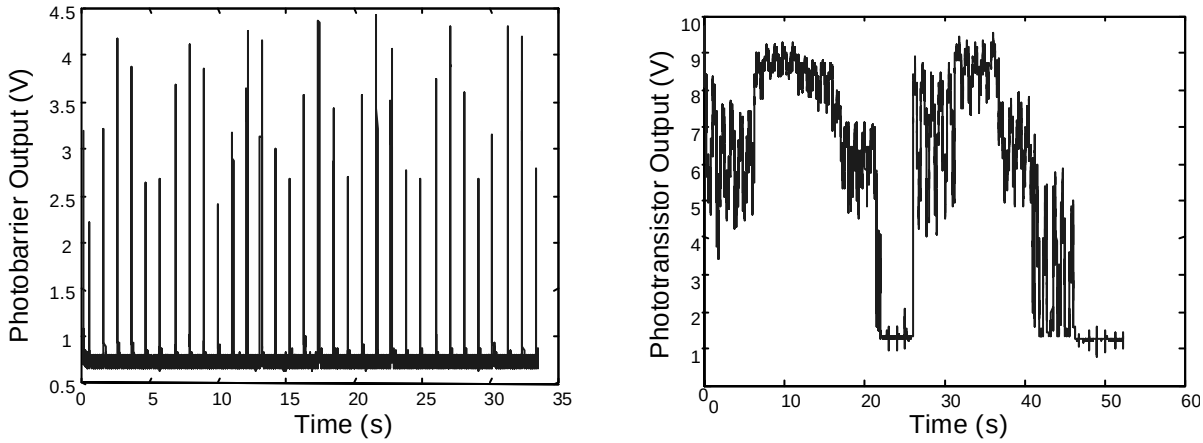
Figure 4. A vessel was simulated as a plastic tube with 5mm of diameter with alternate 2.2cm dark bands.

The phantom was imaged in angiography and fluoroscopy mode during a pullback sequence (Figure 5a and b). A simulated ECG synchronous to the contraction motion was generated with a photoelectric barrier during all acquisitions (Figure 6a). The output of the fiber optic during pullback was detected with a phototransistor coupled with the motor generating the heart-contraction motion (Figure 6b) that provided an output voltage proportional to the amount of detected light with a dynamic range of 8V. This signal is not a two-level digital voltage because the detected light gradually decreases or increases after an opacity change in the vessel. It must be taken into account that the fiber optic gathers light only in the forward direction while the vessel was illuminated obliquely. In consequence, there will be an offset between the position of a change in vessel opacity and the position when this is reflected in the phototransistor output signal. We determined this offset as 5mm in a measurement without respiration and contraction motion. We can observe that the respiration and heart contraction frequency contribute to the noise in the output of the phototransistor. The frequency components corresponding to respiration are filtered out for ulterior calculations. Given the length of the dark bands in the vessel and the time required to advance between them, it was established that the pullback speed was 2.2mm/sec.



(a) (b)

Figure 5. Angiography (a) and pullback fluoroscopy (b) from the heart phantom. The opaque sections of the vessel are marked with partially opaque bands. Radio-opaque marker present on the fibre optic is visible in (b).



(a) (b)

Figure 6. (a) Simulated ECG generated by the photoelectric barrier. (b) Output of the phototransistor connected to the fiber optic.

4. RESULTS

For the experiments, the heart phase was quantized in N=50 steps. For each heart phase, an angiography was selected and a pullback path $\Pi(\delta_n, \beta_n)$ was reconstructed from the respiration-compensated locations of the fiber optic marker according to equations (1) and (2). The respiration compensation algorithm presented by Eck *et. al.*[7] was employed. An example of respiration-compensated fiber optic positions is shown in Figure 7 where the white stars represent the compensated positions. The pullback path was calculated by interpolating between the respiration-compensated fiber optic positions, and taking into account the known pullback speed.

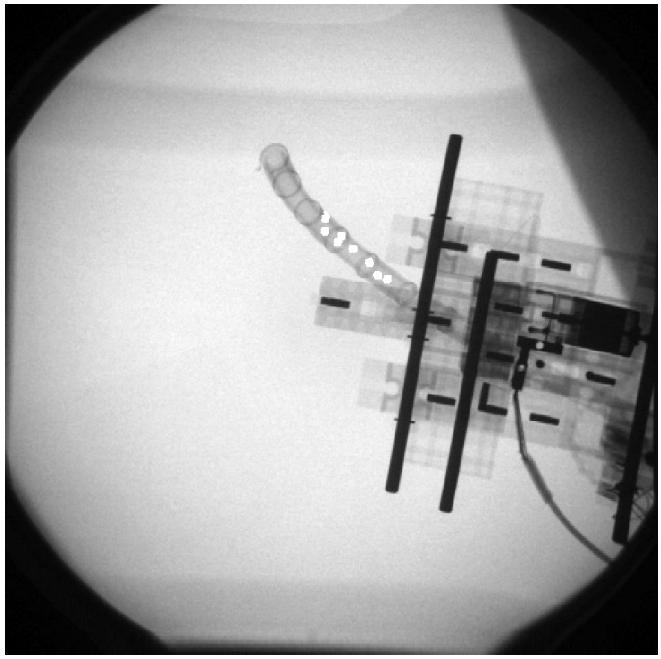


Figure 7. Respiration-compensated positions of the fiber optic. From these points the pullback path is reconstructed by interpolation taking into account the constant pullback speed.

To check the accuracy of the respiration-compensated positions, we manually measured the relative distance of the fiber optic marker to the most distal band marker in the fluoroscopic pullback frames and in the calculated pullback path after respiration compensation. The absolute error for each compensated position in Figure 7 is shown in Figure 8. We can observe that the absolute error is below 2mm. This measurement was performed with three different angiography and fluoroscopy pullback pairs and in all cases the absolute error was below 2mm. The compensation of the respiration motion was identified as major error source in this experiment.

The output of the fiber optic for each position (see Figure 9) was manually compared to the respiration-compensated location. In Figure 9, the shaded areas represent the frames for which the output of the phototransistor should be high because they correspond to an opaque area in the tube. Establishing a threshold of 7V to detect opaque (output higher than 7V) and transparent (output lower than 7V) areas we can observe a time offset of approximately 30 frames, which corresponds to 5.28mm at a pullback speed of 2.2mm/sec and frame rate of 12.5 frames per second. This offset is consistent with the measure that we obtained in the comparison methods without respiration and contraction motion. Consequently, the motion compensation was able to determine a correct correspondence to extracted positions from the x-ray position and the intravascular acquisition.

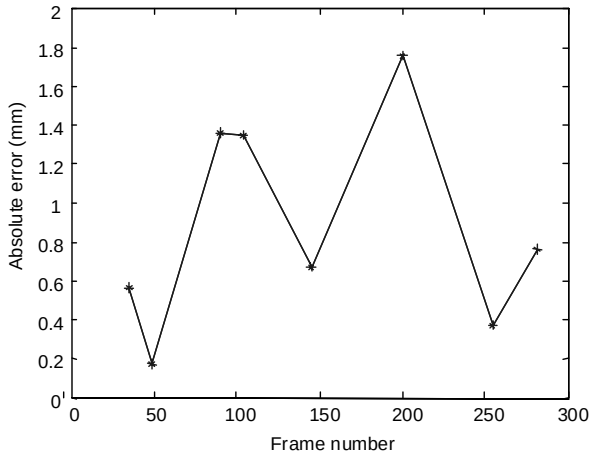


Figure 8. Absolute error between the relative distance of the fiber optic marker in the fluoroscopy pullback frames and its respiration-compensated position in the pullback path.

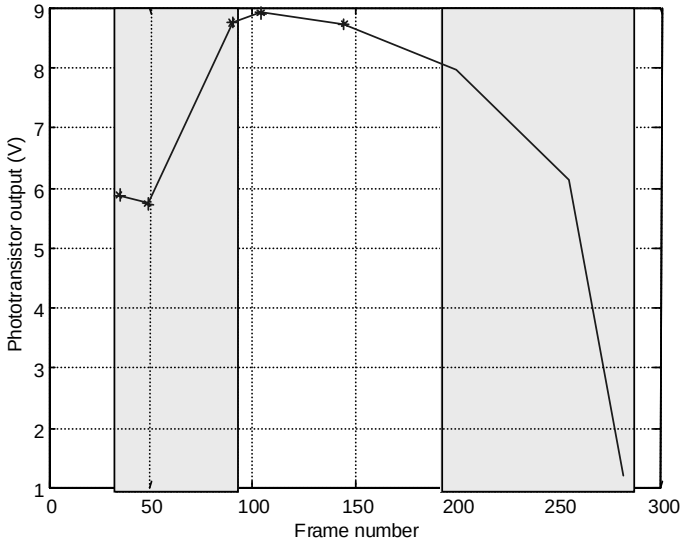


Figure 9. Output signal of the phototransistor at the locations marked in Figure 7. The shaded stripes represent opaque areas in the simulated vessel.

5. CONCLUSIONS AND FUTURE WORK

A method for fusing Cathlab x-ray and IVUS is proposed and demonstrated using a heart phantom. The presented method provides enhanced diagnosis and treatment by registering the acquired IVUS images to angiography during diagnosis and to live fluoroscopy during the intervention. Our measurements indicate that it is possible to reconstruct a two-dimensional pullback path for each quantized heart phase with an accuracy of less than 2mm using a phantom that provides unambiguous position information in x-ray projection and intravascular imaging. Our measurements also indicate that in the phantom case it is possible to register an instrument with a previously acquired IVUS image to an accuracy determined by the respiration compensation algorithm employed. Whereas the contraction phase is determined from the synchronously acquired external ECG signal, the depth of respiration is extracted from the acquired images themselves and, therefore, subject to higher variabilities. In consequence, the method to determine the depth of respiration from projection images has been refined already.[8]

The next important step is the application of the proposed algorithms to patient data to assess the accuracy of the registration methods in clinical application. The accurate motion-compensated reconstruction of the pullback path requires a precise respiration motion estimation. In consequence, the currently used global compensation that is derived from the overall position of the coronaries should be replaced by an estimation of the local motion of coronaries due to respiration.

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