# Intra-operative Simultaneous Catheter and Environment Modelling for Endovascular Navigation Based on Intravascular Ultrasound, Electromagnetic Tracking and Pre-operative Data

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## INTRODUCTION

Cardiovascular diseases (CVD) form the single most common cause of death. Catheter procedures are among the most common surgical interventions used to treat CVD. Due to their minimal access trauma, these procedures extend the range of patients able to receive interventional CVD treatment to age groups dominated by co-morbidity and unacceptable risks for open surgery [1]. The downside associated with minimising access incisions lies at the increased complexity and difficult manipulation of the instruments and anatomical targets, which is mainly caused by the loss of direct access to the anatomy and the poor visualisation of the surgical site. The current clinical approaches to endovascular procedures mainly rely on 2D guidance based on X-ray fluoroscopy, which uses ionising radiation and dangerous contrast agents [2].

In this paper, a Simultaneous Catheter and Environment Modelling (SCEM) method is presented for endovascular navigation based on intravascular ultrasound (IVUS) imaging, electromagnetic (EM) sensing as well as the vessel structure information provided from the pre-operative CT/MR imaging (see Fig. 1). Thus, radiation dose and contrast agents are avoided. The proposed SCEM intra-operatively recovers the 3D structure of the vasculature together with the pose of the catheter tip, which the knowledge of the interaction between the catheter and its surroundings can be provided. The corresponding uncertainties of both vessel reconstruction and catheter pose can also be computed which is necessary for autonomous robotic catheter navigation. Experimental results using three different phantoms, with different catheter motions and cardiac motions simulated by using a periodic pump demonstrated the accuracy of the vessel reconstruction and the potential clinical value of the proposed SCEM method.

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## MATERIALS AND METHODS

The system setup of the proposed SCEM method can be seen in Fig 1. First, for each IVUS image frame, contour extraction was performed with a radial scan [3] to identify high intensity ultrasound reflections, which provides the intersection of the inner vessel wall and the IVUS plane. Suppose $I$ is the contour extracted from the IVUS image, $E$ is the 6DoF pose reported from the EM sensor on the catheter tip, then the proposed SCEM algorithm can be mathematically formulated as a nonlinear optimisation problem which minimise the following objective function

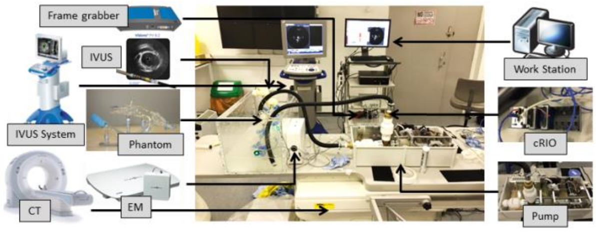


**Fig. 1** The experimental setup.

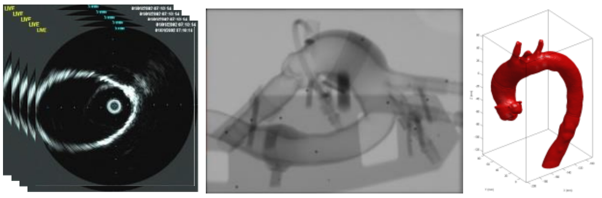

(a) IVUS Images (b) CT Scan (c) 3D Model

**Fig. 2** The IVUS images, the pre-operative CT scan and the aortic model segmented in ITK-SNAP.

$$\text{minimise} \quad F(\mathbf{P}) = \|\mathcal{D}(f(I,\mathbf{P}))\|^2_{\Sigma_I^{-1}} + \|E-\mathbf{P}\|^2_{\Sigma_E^{-1}}$$

where, the state vector $\mathbf{P}$ is the catheter pose, function $f(\cdot)$ transforms the IVUS contour $I$ to the global coordinate frame by using $\mathbf{P}$ and $\mathcal{D}(\cdot)$ is the signed distance function pre-computed from the pre-operative data (see Fig. 2), which presents the shortest distances from all the points on contour $f(I,\mathbf{P})$ to the CT model of the vessel in the pre-operative data, with positive inside the model and negative outside the model. Here, both the IVUS contour $I$ and EM pose $E$ are considered as observations with uncertainties (covariance matrices) $\Sigma_I$ and $\Sigma_E$ [4]. And the optimisation problem tries to minimise the distances between the contour $f(I,\mathbf{P})$ computed from IVUS and the pre-operative model, weighted by the inverse of uncertainty of IVUS contour $\Sigma_I^{-1}$, and minimise the difference between the catheter pose $\mathbf{P}$ in the state vector and the pose $E$ reported from the EM sensor, weighted by the inverse of uncertainty of EM sensor $\Sigma_E^{-1}$. We assume there is no common feature between different IVUS contours, with known EM-CT registration, optimisation using a single frame is

(a) Rigid Phantom (b) Soft Phantom

(c) Silicone Phantom (d) Periodic Pump

**Fig. 3** Aortic phantoms used in the experiments: Rigid phantom made of Plexiglas material (a), Soft phantom made of HeartPrint® material (b) and phantom made by silicone (c) which is compatible with the pump (d) to simulate the periodic cardiac motion.

equivalent to batch/incremental optimisation using a sequence of IVUS images and EM poses.

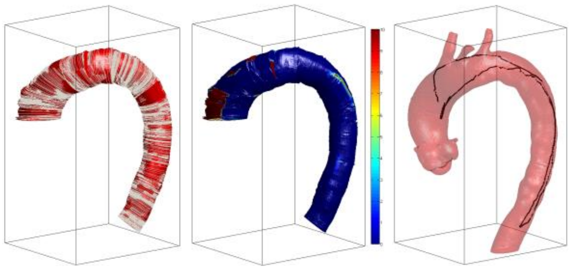
The proposed nonlinear optimisation can be solved using the Gauss-Newton method by linearising the first term in the objective function via Taylor Series, and the state vector **P** is initialised by the EM pose $E$. After an optimal solution of the catheter pose is obtained, the 3D vessel can be reconstructed by using it to transform the IVUS contour $I$ to the global coordinate frame by $f(\cdot)$. The corresponding uncertainty of the catheter pose can be obtained as the inverse of the normal equation matrix evaluated at the optimal solution from the optimisation, and the uncertainty of the vessel reconstruction can be computed by using the nonlinear transformation of multivariate Gaussian distributions [5]

$$\Sigma_P^{-1} = \frac{\partial f^T}{\partial \mathbf{P}} \frac{\partial \mathcal{D}^T}{\partial f} \Sigma_I^{-1} \frac{\partial \mathcal{D}}{\partial f} \frac{\partial f}{\partial \mathbf{P}} + \Sigma_E^{-1}, \quad \Sigma_V = \frac{\partial f}{\partial I} \Sigma_I \frac{\partial f^T}{\partial I} + \frac{\partial f}{\partial \mathbf{P}} \Sigma_P \frac{\partial f^T}{\partial \mathbf{P}}.$$

## RESULTS

In the experiments, three aortic phantoms created by Materialise (Leuven, Belgium) were used (see Fig. 3), which are made of Plexiglas (Rigid), their signature HeartPrint® Flex material (Soft) and silicone (Silicone) respectively. Segmented CT scans of the phantoms provided the triangular surface meshes of the models (see Fig. 2). An Aurora 6 DoF EM sensor (NDI, Waterloo, Canada) was attached to the tip of an IVUS catheter (Volcano, San Diego, USA) to provide its position and orientation.

Insertions and/or pullbacks within the Rigid, Soft and Silicone phantoms were performed to validate the proposed SCEM method. Sudden motions were also performed with the Soft and Silicone phantoms to validate the robustness of the abrupt catheter motions. To simulate the cardiac motion, pump with periodic motions was connect to the pump compatible Silicone phantom, and the aortic phantom was deformed periodically by controlling the frequency and volume per stroke of the pump. The pump signal was then used as the ECG signal to gate the IVUS images in the proposed SCEM method.



(a) Reconstruction (b) Uncertainty (c) Trajectory

**Fig. 4** Results using the soft phantom: 3D reconstruction (a), uncertainty (log) map (b) and the trajectory of the catheter (c).

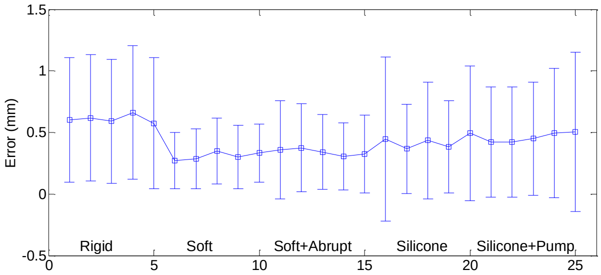


**Fig. 5** Vessel reconstruction error of different aortic phantoms and with abrupt catheter motion and cardiac motion simulated by the pump. (The higher error in the Rigid phantom is caused by lower quality IVUS imaging.)

For each situation, 5 experiments were performed, and the result of one experiment using the Soft phantom is shown in Fig 4. The quantitative evaluation of the 3D vessel reconstruction for all the experiments is shown in Fig 5. The proposed SCEM method runs in real-time as 800 frames per second on an Intel® Xeon(R) E5-2620 CPU @ 2.0 GHz, and the accuracy of the vessel reconstruction was approximately 0.6mm for the Rigid phantom, 0.3mm for the Soft phantom and 0.4mm for the Silicone phantom.

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