Practical Dry Calibration With Medium Adaptation For Fluid-Immersed Endoscopy

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INTRODUCTION

A few endoscopic procedures are performed with a fluid-immersed endoscope. Fetoscopy is one such minimally invasive procedure which allows observation and intervention within the amniotic sac during pregnancy. The fetoscope is inserted through the uterus and is immersed in amniotic fluid. Fluid has a strong influence on the image formation process due to refraction at the interface of the fetoscopic lens which is determined by the optical properties of the amniotic medium (Fig. 1).

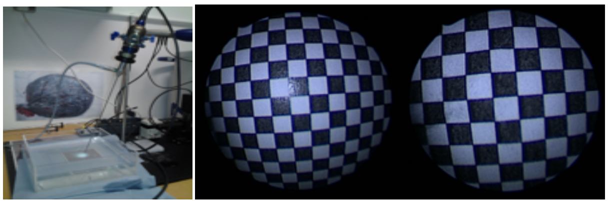

Fig. 1 Images of a calibration target acquired with a fetoscope. Left image: calibration setup. Right image: images of the calibration target respectively in the air and in water.

Accurate calibration is critical to vision-based methods for providing image-guided surgery and real-time information from the surgical site [1]. It consists of recording images of a calibration target of known geometric pattern in order to estimate optical properties of a camera. In the case of a fetoscope, calibration should be realised in the amniotic fluid which cannot be practically realised. A fluid-immersed pre-intervention calibration is also impractical for sterilisation purposes. Few computer vision methods address this issue and most of them are not adapted to the wide field of view as well as the severe distortion effect of a fetoscope [2].
We experimentally show a direct link between dry and fluid-immersed camera parameters which can compensate for the optical properties of bodily or amniotic fluid as well as radial distortion effects.

ENDOSCOPIC CAMERA MODEL

Endoscopic cameras generally embed large field of view optical systems. We chose to rely on the unifying model of central catadioptric system proposed in [3] to handle severe distortion effects and assume optical parameters remain fixed. This model unifies under a common mathematical formalism central catadioptric optical systems with a single projection centre. It has also been proved that the unified model can suit perspective as well as certain fish-eye optical systems which makes it flexible.
The image formation model, illustrated in figure 2, consists of four mappings. A projective ray joins a 3D point Q of the scene with the effective projection center O intersecting the unit sphere in a single point Q_S. The second mapping is equivalent to projecting the point Q_S into the normalized plane from a novel projection center $O'=(0,0,-\xi)^T$. This non-linear mapping intrinsically handles severe radial and tangential distortions. However, a fourth degree function based on a Brown-Conrady model was considered to compensate for lens aberration and model approximations. The last transformation maps the distorted point q_d to the point q observed in the image plane. It is expressed by the internal camera parameter matrix K which embeds the parameter γ (directly related to the focal length f of the camera), a skew parameter (pixel ratio) and the position of the principal point in the image plane. It is worth to note that γ is directly related to ξ according to the real shape of the omnidirectional camera considered [4].

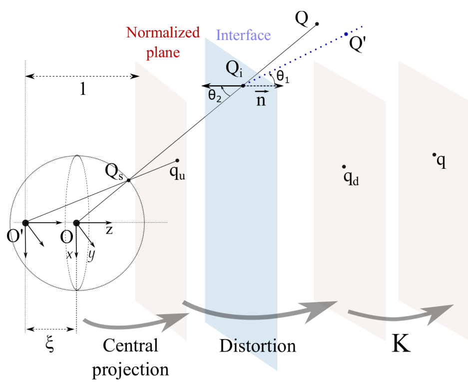

Fig. 2 Unifying model of central catadioptric system.

We now consider a thin interface, planar and fronto parallel to the image plane, which separates the optical system of the endoscope from the external medium (as it appears to be in practice). Because it is difficult to accurately calibrate the depth of the interface we arbitrarily fixed it above the normalized plane without loss of generality. According to Snell's law the refraction angle of the incoming light ray is expressed by:

$$n_1 \sin\theta_1 = n_2 \sin\theta_2 \quad (1)$$

Where n_1, n_2 represent the refractive index of each medium at either end of the interface and θ_1,θ_2 represent the angles between the incoming ray and the normal $\vec{n}$ to the interface. We approximate the refractive index of the amniotic fluid by the refractive index of water n_1=1.33. According to the unified model previously presented, we expect to estimate a new value ξ' and a new focal length γ' to compensate for the refractive properties of fluid. The new value ξ' is obtained thanks to:

$$\xi' = \underset{\xi}{\operatorname{argmin}} \sqrt{(q_{ux} - \frac{Q'_x}{Q'_z + \|Q'\|\xi})^2 + (q_{uy} - \frac{Q'_y}{Q'_z + \|Q'\|\xi})^2} \quad (2)$$

where q_u corresponds to the coordinates of a 2D point on the normalized plane and Q' is an immersed 3D point casting on the normalized sphere on point Q_S. According to Snell's law (1), Q' lies on the light ray belonging to the plane defined by $(\overrightarrow{Q_i, Q_S})$ and $\vec{n}$ and forming an angle θ_2 with $\vec{n}$ at Q_i.
The unifying model of central catadioptric system allows to derive the value of γ' according to ξ' and the shape of the catadioptric mirror. This relationship cannot be applied to a fetoscope which does not embed a real catadioptric system. However, our experiments, presented in the last section, show that an appropriate value of γ' can compensate for the refraction effects without the need to estimate other camera parameters.

METHOD

The calibration method can be decomposed in the following steps. We first realise a dry calibration of the fetoscope using [4]. We use one of the calibration image to estimate the light paths inside the optical system from each pixel of the image plane to the 3D points Qi belonging to the interface plane. We then use equation (1) to compute new incoming light paths considering the refractive index of the external fluid. We randomly define 3D point Q' lying on each of these light rays. Finally we use equation (2) to estimate the new value of ξ'. The back ray tracing algorithm require to convert each mapping of the image formation model. However, the distortion function is not directly invertible. We used [5] to estimate the inverse distortion function.

RESULTS

We conduct experiments with three Storz® fetoscopes model 26120a (33-degree angle of view), 27020aa and 26003aga (0-degree angle of view). Each dataset is composed of 15 dry and 15 underwater calibration images (Fig. 1). We here compare underwater calibration results with the results obtained by estimating fluid-immersed camera parameters form a dry calibration. We used previously defined algorithm to compute ξ'. As the value γ' could not be directly inferred (see section II), we used a single underwater observation to compute it by minimising the reprojection error for the last mapping of the image formation process. We used images realised underwater to compare the two methods. Results are illustrated in figure 3 and synthesised in the following table

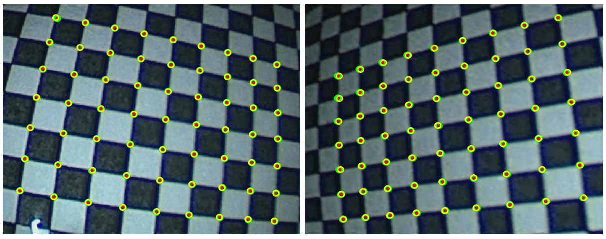

Fig. 3 Reprojection error. Red dots correspond to ground truth. Yellow and green circles correspond to 3D points reprojected in the images using underwater calibration parameters and estimated camera parameters respectively. Left image: 33 degree fetoscope. Right image: 0 degree fetoscope.

Reprojection error (pixel)	Fetoscope 1 (0 degree)	Fetoscope 2 (33 degree)	Fetoscope 3 (0 degree)
underwater calibration	**0.46 ± 0.06**	**0.43 ± 0.43**	**0.52 ± 0.21**
Estimated calibration	**0.69 ± 0.14**	**0.70 ± 0.12**	**0.75 ± 0.12**

The results show that estimated calibration is less accurate than a proper underwater calibration but the reproduction error is less than a pixel for all the fetoscope models used for experiments. Moreover, as most of the distortion effects are corrected underwater, a proper calibration of the fetoscopes using a central catadioptric model was difficult to obtain [3].

DISCUSSION

We presented promising results which show that fluid-immersed endoscope parameters could be directly inferred from a dry calibration using the unifying catadioptric model. Investigations will be conducted in order to establish a proper link between ξ and γ. This could lead to a fast and accurate calibration process adapted to the constraints of fluid immersed endoscopic procedures.

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