Extending Fast Marching Method under Point Light Source Illumination and Perspective Projection

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Abstract

An endoscope is a medical instrument that acquires images inside the human body. An endoscope carries its own light source. Classic shape-from-shading can be used to recover the 3-D shape of objects in view. Recent implementations have used the Fast Marching Method (FMM). Previous FMM approaches recover 3-D shape under assumptions of parallel light source illumination and orthographic projection. This paper extends the FMM approach to recover the 3-D shape under more realistic conditions of endoscopy, namely nearby point light source illumination and perspective projection. The new approach is demonstrated through experiment and is seen to improve performance.

1. Introduction

Endoscopy allows medical practioners to observe the interior of hollow organs and other body cavities in a minimally invasive way. Sometimes, diagnosis requires assessment of the 3-D shape of the observed tissue. For example, the pathological condition of a polyp often is related to its geometrical shape. Medicine is an important area of application of computer vision technology. Specialized endoscopes with a laser light beam head [4] or with two cameras mounted in the head [3] have been developed. Here, we consider a general purpose endoscope, of the sort still most widely used in medical practice.

The problem considered is the recovery of the 3-D shape of tissue in view. Stereo based endoscopy is one approach [6]. The challenge with stereo endoscopy is to determine corresponding features in the two images while the shape of internal organs itself is changing. With a single camera endoscope, shape from shading can be applied. The Fast Marching Method (FMM) [2] is a useful approach. In [2], Lambertian reflectance under parallel light source illumination and orthographic projection is assumed. However, these assumptions are not correct for endoscopy because the light source is nearby and image projection is perspective. In the context of endoscopy, non-Lambertian reflectance, nearby point light source and perspective projection were considered in [5] in which the shape was recovered by propagating the equal distance contours based on the level set method. Extending the FMM for the case of perspective projection is described in [7]. That work still assumes a parallel light source. This paper further extends the FMM for both point light source illumination and perspective projection. The current work is applied to endoscope images. We demonstrate that solving the resulting (eikonal) image irradiance equation with the new method gives better performance via both simulation and experiments on real data.

2. Fast Marching Method

FMM [2] is one approach to solve the eikonal equation fast. Here, the light source direction vector, $\boldsymbol{s}$, and the viewing direction vector, $\boldsymbol{v}$, at each surface point are

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the same and denoted as $(0,0,1)$. For the Lambertian reflectance, the image irradiance, E, is given by

$$E = C(\boldsymbol{s} \cdot \boldsymbol{n}) = C\frac{1}{\sqrt{p^2+q^2+1}} \quad (1)$$

where C is a reflectance factor, and the gradient, (p,q), is $(\frac{\partial Z}{\partial X}, \frac{\partial Z}{\partial Y})$. Eq.(1) can be rewritten as the eikonal equation

$$\sqrt{p^2+q^2} = \sqrt{\frac{C^2}{E^2}-1} \quad (2)$$

Solving Eq.(2) for Z recovers the shape of the object in view. The algorithm chosen to solve the eikonal equation is the FMM [2]. The FMM is summarized as follows:

Step 1 (Initialization). All pixels are labeled as one of three lists, $known, trial, far$, according to the following processes:

1. First pixel is added to $known$ list. Z is assigned to 0.
2. four nearest neighboring points not $known$ are labeled as $trial$ and Z is assigned to f_{ij}.
3. Other pixels are added to far list. Z is assigned as ∞.

Step 2. Select a pixel (i_{min}, j_{min}) with the minimum value of Z among $trial$ list and remove the pixel from $trial$ list and add it to $known$ list.

Step 3. Pixels which belong to far list among four neighboring points around (i_{min}, j_{min}) are added to $trial$ list.

Step 4. Z of the nearest neighboring points of pixel (i_{min}, j_{min}), which belongs to $trial$ list, is calculated and registered.

Step 5. If the $trial$ list is empty, exit the procedure. Otherwise, return to Step 2.

Note that Eq.(2) still embodies the assumptions of parallel light source and orthographic projection.

3. Extension of FMM to Point Light Source and Perspective Projection

Here, the updating equation for Z is derived for imaging conditions appropriate for endoscopy, namely nearby point source illumination and perspective, and the FMM algorithm is modified accordingly. Fig.1 shows the observation system under the condition that both the point light source and viewpoint (center of the lens) are co-located at the coordinate system origin.

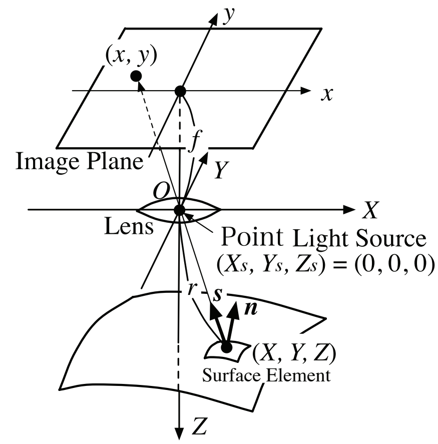


Figure 1. Observation System

3.1. Updating Z under Point Light Source and Perspective Projection

With the observation system of Fig.1, the image irradiance, E, becomes

$$\begin{aligned} E &= C\frac{(\boldsymbol{s}\cdot\boldsymbol{n})}{r^2} \\ &= C\frac{pl+qm-n}{(l^2+m^2+n^2)^{\frac{3}{2}}(p^2+q^2+1)^{\frac{1}{2}}} \quad (3) \\ & r^2 = l^2+m^2+n^2 \\ & l = X_s - X, \quad m = Y_s - Y, \quad n = Z_s - Z \end{aligned}$$

In Eq.(3), r represents the distance between the point light source and the surface point. (l,m,n) represents the difference between the position, (X_s, Y_s, Z_s), of the point light source and the position, (X,Y,Z), of the surface point. When the point light source and viewpoint are both located at the origin, $(0,0,0)$, Eq.(3) simplifies to

$$E = C\frac{-pX-qY+Z}{(X^2+Y^2+Z^2)^{\frac{3}{2}}(p^2+q^2+1)^{\frac{1}{2}}} \quad (4)$$

X and Y can be expressed in terms of Z as

$$X = \frac{x}{f}Z, \qquad Y = \frac{y}{f}Z \quad (5)$$

where (x,y) represents the corresponding image coordinates of the surface point (X,Y,Z). Substituting Eq.(5) into Eq.(4) and rearranging gives

$$E = C\frac{-p\frac{x}{f}Z - q\frac{y}{f}Z + Z}{\left\{\left(\frac{x}{f}Z\right)^2 + \left(\frac{y}{f}Z\right)^2 + Z^2\right\}^{\frac{3}{2}}(p^2+q^2+1)^{\frac{1}{2}}} \quad (6)$$

$$= C\frac{f^2(-px-qy+f)}{(x^2+y^2+f^2)^{\frac{3}{2}}\,Z^2(p^2+q^2+1)^{\frac{1}{2}}} \quad (7)$$

To further simplify, Eq.(7) is rewritten as

$$E = CV \frac{(-px - qy + f)}{Z^2(p^2 + q^2 + 1)^{\frac{1}{2}}} \tag{8}$$

$$\text{where} \quad V = \frac{f^2}{(x^2 + y^2 + f^2)^{\frac{3}{2}}}$$

Solving Eq.(8) for Z gives

$$Z = \sqrt{\frac{CV(-px - qy + f)}{E(p^2 + q^2 + 1)^{\frac{1}{2}}}} \tag{9}$$

where the positive root for Z is selected, based on constraints of the observation system. In Eq.(9), the gradient, (p, q), at an arbitrary point is unknown, in general. But, given the assumption of a smooth sruface, the value at any neighboring point labeled as "$known$" can be used as a good approximation.

The value of (p, q) at any point is estimated from the numerical difference of the Z values between the point and neighboring points. The estimated (p, q) is used in Eq.(9) inside the FMM algorithm to solve for Z of the following point. Updating the list of $known$, $trial$ and far for the surrounding points eventually estimates the shape of the entire target region.

3.2. Initial Values of (Z, p, q) for Initial Points

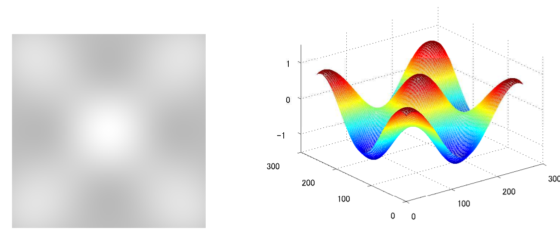
Some initial values are used as initial points for the FMM algorithm. Under parallel light source and orthographic projection, it is impossible to assign different Z values (i.e., heights) to different initial points [2] since image irradiance, E, depends only on the gradient, (p, q), not on Z. On the other hand, under point light source and perspective projection, it is possible to assign initial Z values to some initial points because the inverse square law for illuminance holds and image irradiance, E, depends not only on (p, q) but also on Z at each point. Our method uses the property that $\boldsymbol{n} = \boldsymbol{s}$ at the local brightest point under Lambertian reflectance. The proof for $\boldsymbol{n} = \boldsymbol{s}$ at the local brightest point is given at [1]. Using the above property, the initial values of (p, q) at an initial point are given by

$$p = \frac{x}{f}, \qquad q = \frac{y}{f} \tag{10}$$

Substituting this value of (p, q) into Eq.(9) gives the initial value of Z. This automatic determination for the initial value of Z for initial points is an advantage in comparison with both [5] and [7].

4. Experiments

Our proposed method is compared with [2] (FMM with the parallel light source and orthographic projection) and [7] (a revision of FMM with parallel light source and perspective projection).



(a) Image (b) Shape

Figure 2. 2-D Basis Function

Table 1. Average Error [mm] for Z

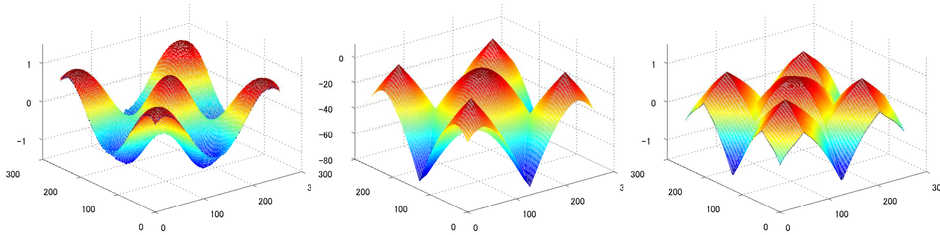
Proposed Method	Paper [2]	Paper [7]
0.1379	26.98	0.4266

Performance first is evaluated by computer simulation. A test image was synthesized using the 2-D basis function given in Eq.(11). The image is shown in Fig.2. For the simulation, the focal length of the camera lens, f, is 10[mm], the rectangle size of image plane is 10[mm] and Z_0 is 15[mm].

$$Z = Z_0 - \cos(X)\cos(Y) \tag{11}$$

Comparison results for the synthesized image (Fig.2) are shown in Fig.3 and Table 1. After selecting the initial points, the initial values of Z determined by our method also were used as initial values for the other two methods. The estimated shape from [2] is different from the true shape. Our method gives a qualitatively better shape with significanntly smaller quantitative error. The result of [7] is similar to that of [2]. The method in [7], like [2], assumes the light source direction is the same for all points (owing to the parallel light source illumination assumption). This introduces some error into the recovered shape. Including initial values for Z at initial points, in addition to values for (p, q), is a distinct advantage of our method.

Finally, performance is evaluated on real data. A real image, acquired with an OLYMPUS GIF-H260 gastroscope, is shown in Fig.4. The original image has 8-bit RGB values. The gray scale image is generated by using the largest dynamic range channel among the RGB channels. The cropped region including the polyp was analyzed by each method. The results are shown in Fig.5. The methods of [2] and [7] recover shapes that are qualitatively incorrect and unacceptable in the application. Our method recovers a shape for the polyp that is qualitatively correct and acceptable in the application.



(a) Proposed Method (b) Paper [2] (c) Paper [7]

Figure 3. Recovered Results for Simulation Image

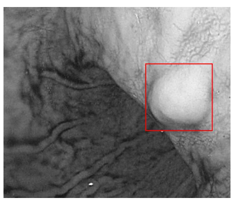
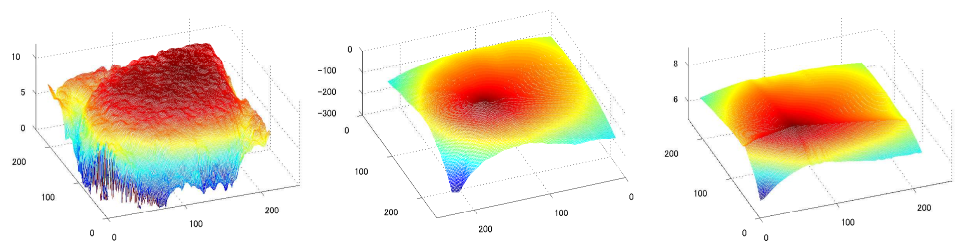


Figure 4. Endoscope Image



(a) Proposed Method (b) Paper [2] (c) Paper [7]

Figure 5. Recovered Results for Endoscope Image

5. Conclusion

This paper described a method that extends previous FMM approaches to shape from shading to include point light source illumination near the camera and perspective projection. Effectiveness is demonstrated by computer simulation and by experiment with a real endoscope image. The result is robust in the case of endoscope images where the distance between camera, light source and object surface is small. The method described still assumes Lambertian reflectance. Actual endoscope images often include a specular component owing to the human body being a "wet" environment. Detecting and removing specular components of reflectance is a potentially useful extension that remains as future work.

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