Optical design and evaluation of a 4 mm cost-effective ultra-high-definition arthroscope

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Abstract: High definition and magnification rigid endoscope plays an important role in modern minimally invasive medical surgery and diagnosis. In this paper, we present the design and evaluation methods of a high definition rigid endoscope, specifically an arthroscope, with a large depth of field (DOF). The incident heights and exit angles of the sampled rays on the relay lens are controlled during the optimization process to ensure an effective field view (70°) and a normal ray path within the limited lens diameter of 2.7 mm. The lens is set up as a multi-configuration system with two extreme and one middle object distances to cover a large DOF. As a result, an entrance pupil of 0.3 mm is achieved for the first time, to bring the theoretical resolution to 23.1 lps/mm in the object space at a working distance of 20 mm, with the wavelength of 0.532 μm. The modulation transfer function (MTF) curves approach diffraction limit, and the values are all higher than 0.3 at 160 line pairs/mm (lps/mm) in the image space. Meanwhile, stray light caused by total internal reflection on the inner wall of the rod lenses and the objective lens is eliminated. The measured resolution in the object space at a 20 mm working distance is 22.3 lps/mm, and test results show that other performance characteristics also fulfill design requirements. The relay lenses are designed with only one type of the spacer and two types of lenses to greatly reduce the fabrication and assembly cost. The design method has important research and application values for lens systems used in modern minimally invasive medical surgery and industrial non-destructive testing area.

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1. Introduction

A medical endoscope is an elongated optical device used to observe otherwise inaccessible areas within the human body through a small orifice in either a non-invasive or minimally invasive manner [1]. These devices are widely used in the medical field for diagnosis and surgical treatment. Endoscopes can be classified as rigid, flexible, and electronic. Flexible endoscopes have a front end whose viewing direction is changeable and thus have an extensive application space. However, the image performance of flexible endoscopes is not as good as that of a rigid endoscope [2].

Bozzini [3] was the first physician who developed methods of examining body orifices in 1805, and the first effective open tube endoscope was developed in 1853 by Desormeaux [4]. This endoscope was used to examine the urethra and bladder. Since then, many scientists and physicians further improved the endoscopic technology in terms of safety and practicality. However, only in 1986 did Mouret developed a video computer chip that allowed the magnification and projection of images onto television screens. This technique of endoscopic surgery became integrated into the discipline of general surgery [5]. Endoscopic surgery has also been proven to be one of the most successful means of providing minimally invasive surgery to numerous patients across the globe and is now routinely performed in various clinical departments with the help of new generation of medical scopes [6–12]. Indeed, endoscopic imaging is rapidly improving, and many therapeutic devices are being developed.
In recent years, endoscopes have been connected with a video camera and displayed on large and high-definition (HD) screens rather than being directly viewed through the eyepiece. Using this new technology enables better observation and recording of images for further analysis and remote diagnosis. Thus, more details are needed to ensure successful diagnosis and operation. Meanwhile, the resolution of the captured image should be increased to match the screen.

High resolution is the key parameter of modern medical endoscopes. From the perspective of clinicians, the detailed features of tissue and lesions cannot be obtained if the resolution is poor; thus, the clinical significance of the endoscope is lost. The lack of high resolution and depth of field (DOF) makes it difficult to perform delicate dissection and suturing, causing surgeons to be less efficient and make more corrective movements than they would in open surgery. HD endoscopes are obviously superior. Kiesslich et al. [13] concluded that HD endoscopes improve small bowel examination. Gross et al. [14] indicated that HD does not enable higher detection rate of advanced adenomas, but significantly smaller polyps can be detected. Rastogi et al. [15] pointed out that HD can identify more adenomas, especially on the right side of the colon. Vakil et al. [16, 17] showed that high-resolution and high-magnification endoscopic examination allows better description of laryngeal lesions with easy characterization of the aspect of neighboring laryngeal mucosa, especially in pre-malignant lesions.

From a technical point of view, display screens are becoming much bigger, which requires higher magnification of endoscopes. Thus, with the resolution of imaging sensors and displays becoming increasingly higher, developing high-magnification and high-resolution endoscopes is necessary. Unfortunately, such high-resolution sensor and small pixels far exceed the imaging ability of a traditional objective-relay lens system, especially with a limited lens diameter. Several optical systems have been developed to meet this high-resolution requirement. However, due to lack of advanced optimization methods, most of these endoscopes have poor image quality, smaller field of view (FOV), or shallow DOF.

Although rigid endoscopes have advanced and been successfully applied in various medical procedures, they still have some limitations that we attempt to ameliorate in this paper. In addition, details on the design methods of a cost-effective and high-performance rigid endoscope are seldom discussed. Stray light is another topic rarely mentioned in the design of endoscope systems and is also discussed in this study.

This paper presents the design and implementation of an HD endoscope, specifically an arthroscope. Section 1 provides a brief history review of the medical endoscope. Section 2 is devoted to the system overview and design considerations of the rigid endoscope. Section 3 discusses the design strategies, and optimization results are presented in Section 4. Section 5 performs the tolerance analyses. Section 6 examines the stray light problem, and Section 7 presents the prototype along with experimental results.

2. System overview and design considerations

To accommodate the increased demand for high-performance endoscopic imaging, a modern endoscope must generate an image as sharp as possible. The important factors to consider during the design process are listed below.

2.1 High resolution

The factors that limit a system's approach to high resolution include pupil size and residual aberrations. For a medical endoscope, the resolution is defined by [18, 19]

\[ r(d) = \frac{D_e}{1.22\lambda d} \]  

where \( d \) is the working distance, and \( D_e \) is the entrance pupil diameter, \( \lambda \) is the wavelength.

The resolution \( r(d) \) must be proportional to the entrance pupil diameter at a fixed working distance \( d \); thus, the entrance pupil diameter \( D_e \) must be increased to create HD endoscopic
images. However, $D_e$ is limited by the outer diameter of the endoscope lens. As shown in Fig. 1, a rigid endoscope usually consists of an outer tube, an inner tube, lenses, and optical fibers. The lenses are placed inside of the inner tube, whereas the optical fibers are assembled between the outer and inner tubes, taking the charge of illumination. The wall thickness of the outer and inner tubes is 0.1 mm. A 4 mm endoscope usually uses more than 1500 optical fibers, so the physical diameter of the lenses is about 2.8 mm. The clear aperture is about 2.6 mm in consideration of the spacer, and the heights of all the intermediate images in the lens system are smaller than 1.3 mm. While keeping an FOV of 70°, 0.3 mm is the upper limit of $D_e$ for a 4 mm endoscope, and this is determined by Eq. (2). For an endoscope system, the F-number of the objective lens is matched with that of the relay lenses.

$$D_e = \frac{f_{obj}}{f_{relay}}$$  \hspace{1cm} (2)

where $f_{obj}$ is the effective focal length of the objective lens, when the FOV is 70°, $f_{obj} = 1.3/\tan(35°) = 1.85$ mm; $f_{relay}$ is the effective focal length of the relay lenses, and $f_{relay} = 1.3/\tan(θ_t) = 16.09$ mm as shown in Fig. 2; $D_{relay}$ is the pupil diameter of the relay lenses, and its maximum value is 2.6 for a 4 mm endoscope.

Meanwhile, axial aberrations such as spherical aberration are proportional to the third power of $D_e$. Each relay sub-unit should be well corrected from axial aberrations, particularly from spherical aberration and astigmatism. The endoscope system in this paper consists of an objective, an eyepiece, and three sets of relay lenses (see Fig. 8). The individual relay sub-units must have small axial aberrations because the axial aberrations of the complete relay system are equal to those of a single sub-unit multiplied by the number of units. Distortion causes wrong perception but can be corrected by an electronic method and then displayed on an HD screen. Thus, distortion is not a major concern in this design.

2.2 Small field curvature and extended DOF

If the DOF is small or the field of curvature is large, only a small portion of the observing image is clear; thus, the peripheral area of the image is blurry and some details may be lost.
An unclear peripheral area of the lesion may reduce the effective FOV, cause eye fatigue, and result in difficulty in diagnosis and surgery. This factor may also cause injury, especially when the surgical instruments are out of the FOV. Thus, each relay unit should have a flat field, and the entire system should be designed with an extended DOF.

2.3 Telecentricity

One special requirement for objective and relay lenses used in rigid endoscopes is telecentricity [1], which avoids severe vignetting and prevents light loss during image transmission from the distal to the proximal end. If the lens system does not meet the telecentric requirement, some rays fail to transmit to the proximal end. The loss ray may be absorbed by the spacer or may be reflected back to the successive lens by total internal reflection (TIR) if the ray hits the inner wall of the lens, especially the rod lens. These rays are stray light, which reduce image contrast.

2.4 High contrast and elimination of stray light

The photos shown in Fig. 3 are taken through an endoscope targeting a fluorescent lamp. The halos and ghost image around the fluorescent lamp in the photo have been observed to be both in-field type stray light. Thus, the causes of this phenomenon must be figured out, and the stray light should be eliminated to create the best performing system. Ways to control stray light to meet lower levels of stray light demands must be considered during the “preliminary” conceptual design [20]. Otherwise, the stray light would be difficult to suppress after the assembly of the endoscope system. Frequently, stray light analysis is an afterthought, left as something to be dealt with when the rest of the design is complete. This mindset however can be severely detrimental to a project [21]. Thus, consideration and analysis of stray light issues should be conducted during the optical and mechanical design phase of system development.

![Fig. 3. Photo taken through a commercial endoscope when the fluorescent lamp locates at four radial object positions: (a) on-axis, (b) 0.38 field, (c) 0.67 field, and (d) full field.](image)

We modeled an endoscope system in the non-sequential ray tracing software LightTools [22]. The objective lens is shown in Fig. 4(a). To identify and track troublesome light paths, stray light paths and problematic areas where stray light may occur must be identified. Moreover, the magnitude and distribution of scattered light must also be determined. We placed a point source on the object plane and set the targeting zone at the distal end of the endoscope, and a receiver with a ray path filter was placed on the image plane. By ray tracing simulation, all different ray paths are marked in rays with different colors, as shown in Fig. 4(b). Only a few rays incident on the first lens are useful, and some of the rays are blocked by the spacer. Conversely, some rays are further reflected into the successive lens. Figures 4(c) and 4(d) show the magnified front and end area of the system, respectively, which show the reflected light rays from the inside wall of the objective lenses.

Stray light (ghost image) may also be due to double reflections on optical surfaces. However, antireflection effects are achieved even with only a single layer of coating if the vitreous material of the rod lenses in the relay lens system has a high refractive index [23]. This condition also helps reduce the inherent field curvature of the relay lens because the power of all relay groups is positive and all surfaces are spheres [24].
Fig. 4. Deliberately exaggerated example of an endoscope objective lens with long thickness and small air space distance. (a) Objective lens with an imaging (red) ray bundle; (b) Objective lens with a ray bundle that enter the first lens; (c) magnified parts of the objective lens.

Fig. 5. Light distribution on the image plane with a point source at four radial object positions: (a) on-axis, (b) 0.38 field, (c) 0.67 field, and (d) full field. Top row shows the Pseudo-color results and bottom row shows the true color simulation results.

In addition, all lenses and spacers should be designed as simple as possible for easy fabrication and distinguishable assembly.

Table 1 lists the specifications of the HD endoscope. The overall length of the arthroscope is 210mm, the entrance pupil is 0.3 mm, which is the limit of a 4 mm endoscope. The effective focal length of the objective lens is 1.8 mm, which corresponds to an F/# of 6, the full FOV is 70°, and the image diameter is 2.2 mm. However, these requirements make it difficult to achieve a high-performance and wide-angle endoscope design [25, 26].

Table 1. Design specifications of high performance endoscope

| Optical Parameters                  | Requirement |
|-------------------------------------|-------------|
| Direction of view                   | 0°          |
| Full Field of View (FOV)            | 70°         |
| Endoscope Diameter (Φ)              | 4mm         |
| Magnification                       | 0.09        |
| Clear Aperture (C4)                 | 2.6mm       |
| Entrance Pupil Diameter (D)         | 0.3mm       |
| Endoscope Length (L)                | 210mm       |
| Working distance (d)                | 20mm        |
| Wavelength (λ)                      | 486-656nm   |
| Objective lens thickness (mm)       | 0.8~1.5mm   |
| Radius of curvature (mm)            | >2mm        |
3. Design strategies and result analysis

The lens is optimized and evaluated using the commercial optical design software CODEV [27] to achieve diffraction-limited performance. To ensure that the endoscope image is clear in a large DOF, three configurations are set up during the optimization process, with two extreme and one middle observing distances.

A practical approach is to design an endoscope with objective and relay systems together [28, 29]. This method relaxes the flat-field requirement for the subsystem. The relay has under-corrected astigmatism and field curvature that balances the intentional over-correction of the objective. The objective is a reversed telephoto type that is well suited for this kind of overcorrection because of the strong negative element in the front. The rod lens is the best choice for the relay lens system if no vignetting is allowed because it performs better than other relay systems [1, 30]. An important problem when designing relay systems is that all relay units has a positive power; hence, the Petzval sum is always positive, inwardly curving the final image surface. A solution to this problem is to compensate with the proper amount of astigmatism, to seek a flat tangential surface, and/or to reduce the Petzval sum as much as possible using thick elements [29].

We adopt the evolutionary Hopkins rod lens in this design to simplify the fabrication and mounting of lens elements. One relay unit consists of two similar triplets centered exactly on the stop, as shown in Fig. 6. The triplet is extremely thick to reduce the Petzval sum that is not fully corrected using moderately high index crown glass. The triplets use flint glass on the negative element, and the outer surface of the flint is convex so that it can function as the field lens. In a triplet, two end lenses are affixed to the opposite ends of the center lens. Furthermore, the two meniscus lenses on the side of the triplet are physically the same. In other words, the relay system consists of two kinds of lenses, but only three test plates are needed for its fabrication. In addition, all axial distances between any adjacent rod lenses are the same, i.e., only one type of spacer is needed for the assembly of the relay system. Each relay sub-unit transfers the image by more than 60 mm in this design, and the relay as a whole transfers the image by more than 180 mm from the distal to the proximal end. This significantly reduce the fabricating and aligning complexity. However, it greatly increases the design difficulty.

![Fig. 6. Relay lens unit.](image)

3.1 Optimization variables

Apart from the first surface of the objective lens, all available parameters including radius of curvature, glass, and thickness are set as variables, which include 23 variables. The relay lens consists of three sub-units each consisting of rod lens pairs with 7 variables, including two radii of curvature, two thicknesses, and two glasses. For all other surfaces, the thickness and glass are picked up from the first two lenses. The doublet eyepiece has 9 variables, including 3 radii of curvature, 4 thicknesses, and 2 glasses, with a total of 40 variables.

3.2 Intermediate image position constraints

Four intermediate image planes exist in the design, which is a practical disadvantage with the configuration if the outer surfaces of the rod lenses are located exactly at the image plane, and
any surface imperfection and dirt on this lens can be clearly observed on top of the image. Thus, the side surfaces of the rod lenses must be kept away from the image plane during the design process, either ahead of or behind the image, so that imperfections on the side surfaces are out of focus and are not visible. We added intermediate image position constraints to slightly move the field lens away from the intermediate image, but the image distance of the last rod lens is set as a variable to correct the field curvature and spherical aberration. One way to do this process is to add another configuration and set the intermediate plane as the final focal plane. However, this process increases the complexity of the design and slow the optimization process. In this paper, we select two marginal reference rays of the center field on the tangential plane and control the z value along the optical axis of their intersection point, keep it positive with respect to the outer surface where they only leave the rod lens and negative with respect to the adjacent outer surface where they hit.

3.3 Constraints on aberrations

The relay system is completely symmetric, including the object and image distances. Coma, distortion, and magnification chromatic aberrations are automatically cancelled out. Thus, coma correction with lens bending need not be of concern. Axial aberrations such as spherical aberration are controlled by limiting the corresponding third-order aberration.

3.3 Stray light constraints

According to the analysis in Section 2, several constraints should be included for limiting stray light during optimization:

1. A physical stop is added to the design, and the spacer can be coated with a thin layer of flat black paint or chemically blackened. In some designs, no physical stop exists in the objective lens, and the spacers absorb the rays that overflow the limited aperture, thereby functioning as a stop. However, some rays hit the edge of the lenses, and the incident angle is usually larger than the critical angle; consequently, TIR occurs and the TIR condition cannot be broken because it is inside a lens. However, the drawback of this process is that some rays hit the edge of the lens and turn out to stray light.

2. The distance between the first two lenses is controlled. Thus, a longer spacer results in the absorption of more unwanted rays rather than being transmitted into the second lens.

3. The edge thickness of the lenses before the stop is limited; thus, fewer rays hit the edge.

4. The radius of curvature of the front surface of the second lens is also controlled. The negative radius refracts the rays outward to the edge; thus, more rays hit the edge. However, the positive radius refracts the rays inward to the optical axis, thereby avoiding the chance that the rays hit the edge.

5. The telecentric condition is controlled by the chief ray angle on the image plane, which helps reduce the TIR inside the rod lens and improves light-transfer efficiency.

3.4 Glass substitution

An endoscope is often exposed to water, blood, and fluid flow. A rigid endoscope needs to be cleaned to prevent infection through high-temperature sterilization before use. Thus, the glass types used for the front and rear lens surfaces must be selected for high weather and stain resistance.

As part of the design process, the material selection is critical. The lens system must be optimized with fictitious glass to maximize performance and then convert fictitious glass into real glass with the closest index and dispersion. Sometimes, awful designs are created even
for optimized system before conversion. More importantly, optimization with fictitious glass does not discriminate on cost, stain resistance, transmission loss, and glass availability.

The solution is to use CODE V Glass Expert [31] at the final design stage because it examines the user-selected elements of a system and attempts to find the optimum set of user-selected real glass types to maximize system performance. For each iteration, the proprietary algorithm first replaces a material for a lens from the user-selected glass catalog and then performs further optimization with previous constraints. Finally, a decision on whether to keep or reject the result based on optical performance, absorption, cost, weight, and thermal expansion is reached. The index range is limited within 1.4 and 1.85; thus, LA series glasses are avoided for the rod lens. As shown in Fig. 7, the error function (ERF) of the design decreases from 0.956 to 0.5 (almost by half); otherwise, the ERF of the design stagnates at around 1.

![Error Function VS Optimization Time](image)

**Fig. 7.** Glass optimization improvement versus time curves.

### 4. Design results and analysis

The final optical design of the endoscope system after global searching and glass substitution is shown in Fig. 8. The system is optimized at three different object distances to cover a large DOF. The largest object distance for optimization is set at 500 mm, which is much larger than what is needed in real applications.

![Optical layout of the endoscope](image)

**Fig. 8.** Optical layout of the endoscope, wherein three sets of relay lens pairs are disposed between an objective lens and an eye lens.
Fig. 9. Transverse ray aberrations for the 0.0, 0.38, 0.67, and 1.0 field of the three working distance (WD). (a) WD = 20 mm; (b) WD = 10 mm; (c) WD = 500 mm.

Figures 9, 10, and 11 show that the system as a whole is diffraction limited, with small residuals of axial color and spherical aberration. The transverse ray aberration plots in Fig. 9 indicate that spherical aberration and lateral chromatic aberration are all present in small amounts. Figure 10 shows the spot diagram of the three configurations, the RMS spot sizes are all smaller than the Airy disk. Figure 11 shows that the design has good contrast over most of the field at 160 lps/mm, the MTF values are all above 0.2; thus, it should work well with the current generation of high-resolution CCD sensors having 3–5 µm pixel.

Fig. 10. Geometric spot diagrams for four radial image positions: on-axis, 0.4, 0.7, and 1.0 field of the three configurations. (a) WD = 20 mm; (b) WD = 10 mm; (c) WD = 500 mm.
Fig. 11. Polychromatic modulation transfer function curves of the systems with different WDs for four radial image positions: on-axis, 0.38, 0.67, and 1.0 field. (a) WD = 20 mm; (b) WD = 10 mm; (c) WD = 500 mm.

Longitudinal spherical aberration, field curvature, and distortion are computed at wavelengths of 486, 587, and 656 nm as a function of field angle in Fig. 11. Field curvature and astigmatism are almost the same for different DOFs. The relay has no distortion because of symmetry; thus, the distortion of the system is equal to the distortion of the objective. The distortion is left uncorrected in this design, as is customary in endoscope systems, in which the maximum value is −19.8%, which will be rectified during image processing.
Fig. 12. Distortion plot in percentage of deviation from chief ray location of the systems with different WDs (a) WD = 20 mm; (b) WD = 10 mm; (c) WD = 500 mm.

Fig. 13. (a) The 11th pattern of resolution test board (JBT 9328-1999 A3) [32]; (b) PSF plots of the center field; (c) simulation image of the center field; (d) PSF plots of the marginal field; (e) simulation image of the marginal field.

Figure 13 shows the image simulation results in CODE V, the input image is a custom JBT 9328-1999 A3 resolution board [32]. Figure 13(b) shows the PSF plots of the center field, Fig. 13(c) is the simulation image of the center field (within ± 5°); while Fig. 13(d) shows the PSF plots of the marginal field (within ± 5°) and Fig. 13(e) is the simulation image of the marginal field. Figure 14 shows the image simulation results of RGB circle image, the color restoration of the system is good.
Fig. 14. Image simulation result of the color image. (a) The original RGB circle image; (b) the simulated image.

Fig. 15. Relative illumination curve of the final design.

The relative illumination (RI) curve shown in Fig. 15 is computed as the ratio of corner luminance to the center luminance. All images from the endoscope vary in intensity from the center to the edge, the image center is always the brightest portion of the image for most of the designs, but in this design, the peak value happens at the field of 26°, this helps to balance the RI across the field, the RI value of the marginal field is over 94%. This is due to the large field of view and severe pupil aberrations – similar to a fisheye lens.

5. Tolerance analysis

To ensure successful prototype fabrication, tolerance analysis is performed using the criterion for the maximum allowable decrease of 0.01 in MTF value at 120 lps/mm. For comparison, the nominal MTF of the design is diffraction limit. Using CODEV to analyze the symmetric cross of the sagittal and tangential field positions, tolerances of the optical component fabrication and alignment are specified and displayed in Table 2, the shift and tilt angle of the image plane are used as the compensators. Monte Carlo analysis is also performed using CODEV to generate 1000 perturbed optical systems. The mean MTF for the trials is 0.4, and >80% of the trials have an MTF larger than 0.38 at 120 lps/mm, as shown in Fig. 16.

Using standard definitions of tolerance levels [26], the majority of tolerances in Table 2 fall under either commercial or precision tolerance requirements, with only the decenters on surface 2, lens 2, and lens 3 requiring high-precision manufacturing and assembly. Figure 16(b) shows the as-built MTF plots of the system, it predicts the MTF performance of the system with 2-sigma probable change for the current set tolerances and compensators, the MTFs of all fields at 120 lps/mm are over 0.2, the value of the center fields are higher than 0.3.
Fig. 16. (a) Cumulative possibility estimates and (b) as-built MTF plots of the tolerance analysis at four fields for the endoscope lens optimized for best nominal performance.

Table 2. Tolerances value of the endoscope system.

| Tolerance Item    | Value                                         |
|-------------------|-----------------------------------------------|
| Radius            | ± 1%                                          |
| Thickness(mm)     | ± 0.03 for relay lenses ± 0.02 for objective lens |
| Diameter(mm)      | ± 0.05                                        |
| Surface Irreg.    | <3 fringes                                    |
| Surface Decenter(um) | ± 10 for objective lens ± 20 for the relay lens |
| Surface tilt(°)   | ± 20                                          |
| Index tolerances  | 0.002                                         |
| Abbe tolerances   | 0.008                                         |

6. Stray light analysis and examination

In this section, we analyze the stray light of the designed endoscope system and model the lens system with the housing in LightTools. Given that light may illuminate the housing, which may scatter the light onto the focal plane, the optical properties of the housing is set to absorb the light. According to stray light constraints, the first two objective lenses are separated and kept to have small edge thickness. The first spacer is long, and the front surface of the second lens meets the anti-TIR condition, indicating that the rays are converged toward the optical axis rather than to the edge. The three front lenses of the objective lens are shown in Fig. 17. The ray paths through the objective lens are shown in Fig. 18. Most rays enter the first lens, either absorbed by the housing or blocked by the stop, and no ray hits and reflects on the lens edge.

Meanwhile, the objective lens meets the telecentric condition; thus, the rays do not overflow or reflect inside the rod lenses. All optical surfaces are also coated with an anti-reflection coating to decrease reflections on the surfaces to a minimal value. Further analysis with a complete endoscopic system is carried out, and the light distributions of several field
points at the image plane are analyzed and shown in Fig. 19. For a point source on the object plane, only one corresponding image point is on the image plane, and no stray light is observed. Analysis results show that TIR minimally affect the image contrast.

Fig. 18. Endoscope objective lens with imaging and straylight path.

Fig. 19. Light distribution results on the image plane with a point source at four radial object positions: (a) on-axis, (b) 0.38, (c) 0.67, and (d) full field. Top row shows the Pseudo-color results and bottom row shows the true color results.

7. Prototype and experimental results

After fabricating all lenses, we first assemble and test the endoscope on a V-block and then in the tube to test performance, as shown in Fig. 20. We find that the second and third lenses are the most sensitive elements during assembly, consistent with the tolerance results.

Fig. 20. The rod lens and the V-blocks experiment.
The experimental setup for performance testing consists of a collimated light source, a resolution target, the prototyped endoscope, an erector tube and a camera. The setup is shown in Fig. 21. The resolution target used in this study is a custom JBT 9328-1999 A3 resolution board [28]. Figure 22(a) shows the image result relayed through a commercially available endoscope, only the 6th or 7th patterns can be distinguished. However, as shown in the Fig. 22(b), with the endoscope developed in this paper, the 10th and 11th patterns can be clearly distinguished, as shown in Fig. 22(c). According to the size of the 11th pattern shown in Fig. 13, the achievable resolution of the endoscope at the object space is 22.3 lps/mm, which is close to the theoretical value of 23.1 lps/mm calculated using Eq. (1). The result indicates that the resolution of the designed endoscope is favorable compared with previous designs of the same size.

Figure 23 shows the testing image of the FOV. The front end of the endoscope is placed at the center of an arc with a radius of 57.3 mm. According to definition of a radian, an arc length of 1 mm corresponds to 1°. The FOV shown in the Fig. 23(b) is 69.5°, almost the same as the designed value of 70°.

Fig. 22. Photographs obtained from a custom JBT 9328-1999 A3 full resolution target. (a) resolution target at the center field of a commercially available endoscope, (b) resolution target at the center field of the prototyped endoscope, (c) magnified area of the 10th and 11th pattern.
The test image at a nominal working distance of 20 mm is shown in Fig. 24(b), where the details of the center and edge of the image can be clearly observed, i.e., a small field curvature is achieved. To further test the DOF, the position of the target is changed to obtain a series of different objective distances. The images of the target in different distances are shown in Fig. 24. As annotated, the images at all testing distances from 10 mm to 125 mm can be distinguished. Thus, the DOF of the designed endoscope is 115 mm, which is significant for diagnosis. Figure 25 shows the photographs of a point source at four different fields through the developed endoscope, and no ghost and halo image is observed. Figure 26 shows the image of a palm, in which the palm print can be vividly observed.

![Fig. 23. (a) FOV testing experiment setup; (b) test image of the developed endoscope.](image)

![Fig. 24. Test image with different WDs. (a) WD = 10 mm; (b) WD = 20 mm; (c) WD = 30 mm; (d) WD = 50 mm; (e) WD = 70 mm; (f) WD = 125 mm.](image)

![Fig. 25. Images of a point source at four different fields through the developed endoscope.](image)
8. Conclusion

The design and analysis method of a high-performance rigid lens are presented. The causes of stray light are determined by ray path analysis. Results show that suppression methods including the adjustment of the objective structure and control of the ray path are proposed. Corresponding constraints are used for optimization. As an implementation of the aforementioned methods, a rigid endoscope is designed and fabricated with a FOV of 70° and an EPD of 0.3 mm. The MTF approaches the diffraction limit and becomes higher than 0.2 at 160 lps/mm, and the size of the spot diagram is smaller than 4 μm. To the best of our knowledge, this EPD is the largest of this type of endoscope ever made. The measured resolution is 22.3 lps/mm, the DOF is 115 mm and the FOV is 69.5°. In addition, stray light examination indicates that stray light is eliminated in the optical system. The complexity of the fabrication and assembly of the design are markedly reduced by the optical design because the relay system uses only two kinds of lenses and one spacer. The lenses in the objective system are easy to recognize for assembly; for example, the radii on both sides of the double convex lens are the same. Finally, the performance of the prototype is verified, and the designed endoscope is proven capable of acquiring high-resolution and high-contrast in vivo images for diagnosis and surgery.

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