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Coherent Narrow-Band Light Source for Miniature Endoscopes

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***Abstract*—In this work, we report the successful implementation of a coherent narrow-band light source for miniature endoscopy applications. A RGB laser module that provides much higher luminosity than traditional incoherent white light sources is used for illumination, taking advantages of the laser light's high spatial coherence for efficient light coupling. Notably, the narrow spectral band of the laser light sources also enables spectrally resolved imaging, to distinguish certain biological tissues or components. A monochrome CMOS camera is employed to synchronize with the time lapsed RGB laser module illumination for color image acquisition and reconstruction, which provides better spatial resolution than a color CMOS camera of comparable pixel number, in addition to spectral resolving.**

***Index Terms*—Endoscope, monochrome CMOS, narrow-band imaging (NBI), RGB laser**

I. Introduction

A bright and uniform white light illumination is a critical factor in endoscopy [1], [2]. Conventional endoscopic light sources consist of xenon or tungsten halide lamps or light emitting diodes (LEDs), which are either coupled through an optical fiber bundle placed in the endoscope tube or used as a ring shaped assembly in the distal end of the endoscope [2], [3]. Critically, if the footprint of the light source can be greatly reduced (hence the diameter of the endoscope), it will not only decrease patient's discomfort by having a smaller wound but also facilitate unprecedented endoscopic procedures, such as insertion through a needle. Consequently, the risk of infection and the time required for wound recovery/healing can be lowered accordingly.

With the advent of miniature CMOS sensors [4], [5], [6] of outer diameter after packaging as small as 0.8 mm [7] and the emergence of high definition imaging guide (HDIG) [8], the bottleneck in further reducing the total diameter of the endoscope depends critically on the footprint of the light sources. In this context, an LED [9], [3], [10], [11], [12], [13], which has recently become a standard light source for endoscopic diagnosis and surgery, despite the inherent advantages of longer lifetime, higher efficiency and lower power consumption, is unfortunately incoherent and therefore needs a fiber bundle [14], [15] for efficient light coupling because of the large LED dispersion angle. Such fiber bundle based light delivery structures typically possess an outer diameter of several millimeters [16], which ultimately overshadows the advantages of miniature CMOS cameras. Although, packaging LED chips along with the CMOS at the distal end of the endoscopic tube could be an alternative, the required dimension of the printed circuit boards for both the LED chips and the CMOS image sensors prevent miniaturization of the effective diameter of the endoscope [3], [12], [13], [17].

On the other hand, a spatially coherent light source coupled with a downsized light delivery system would greatly facilitate ultrathin endoscopes [18]. In a previous work, we have successfully implemented a supercontinuum laser with a high degree of spatial coherence as the light source for ultrathin endoscopy [19]. However, the inherent disadvantages such as high price, complexity in optimizing the illumination property and bulkiness prohibit supercontinuum from being practical for clinical treatment. In this context, there is an urgent need for a compact and cost effective alternative of supercontinuum laser that could become an ideal source for endoscopic illumination.

The aim of this work is to develop a lighting system that would not only further reduce the footprint of the illumination system for an endoscope (with diameters from 2.7 mm down to 0.49 mm) but also synchronize with the imaging acquisition to achieve narrow band spectral resolution. Here, we use a RGB laser module with a narrow band spectrum (red, 635 nm; green, 532nm and blue, 445 nm) to facilitate user-friendly operation and acquisition of spectral images. The present design allows software control of the illumination duty cycle [20] (i.e. the

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effective overall luminosity) of each of the three primary colors (wavelengths) for white balance [21] and image contrast.

II. Materials and Methods

Our setup is shown in Fig. 1(a). The red, green, and blue lasers are combined with dichroic mirrors into the same light path before coupling into a multimode fiber (NA: 0.50, core diameter: 200 µm, cladding diameter: 225 µm, transmission range: 400 – 2200 nm) for efficient light delivery, where NA refers to numerical aperture. Notably, high spatial coherence of the RGB laser output facilitates highly efficient light coupling and delivery. For effective coupling, a customized pair of collimators, as shown in Fig. 1(b), is used to relay the output of the multimode fiber to the illumination fiber bundle (NA = 0.57) of the ultrathin endoscope. The illumination divergence (~70°) of the fiber bundle meets the FOV (~65°) requirement of image acquisition. Notably the illumination fiber bundle is consisted of 20 fibers, and to guide light from the collimator to the endoscope distal end as shown in 'Inset 1' of Fig. 1.

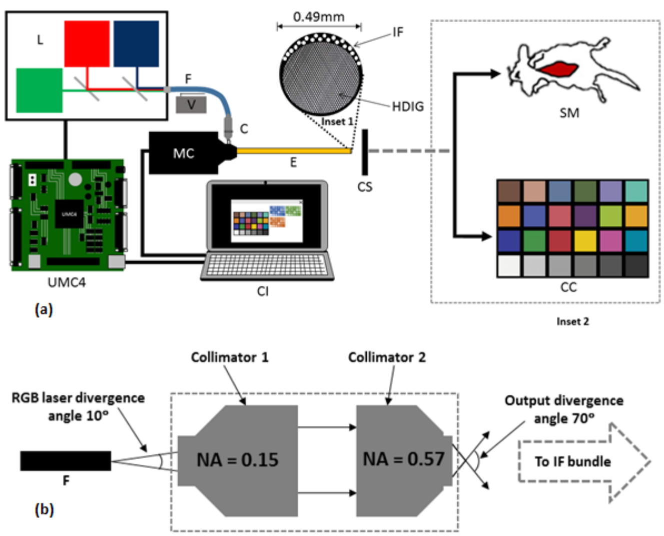

Fig. 1. (a) The schematic of the proposed endoscope system with synchronized laser illumination. L: illumination source (RGB laser); F: multimode optical fiber; V: mechanical vibration; C: collimator with custom design; MC: monochrome CMOS camera; E: endoscope tube; SM: surgical model; CC: color checker; CI: computer interface. Inset: cross-sectional view of the endoscope. HDIG: high definition image guide (10K pixels) with 65° FOV; IF: illumination fibers (the number of fibers is 20) with 70° divergence angle. Inset 1: The cross-sectional view of the endoscope tube containing the 20 IFs and the high definition image guide (HDIG). Inset 2: top view of the surgical model and color checker. (b) Schematic of the customized pair of collimators used to make the illumination divergence angle met the FOV requirement of 65°. Notably the numerical aperture (NA) of the first collimator is 0.15, which covers the laser beam exiting the fiber that has a divergence angle of ~10°. On the other side, the NA of the second collimator is 0.57, to match the NA of the IF bundle, which has a divergence angle greater than the viewing angle of HDIG.

In addition, due to the coherent nature of the laser beams, parasitic speckle is also generated as a side effect, resulting in severe degradation of image quality. To remove the speckle and non-uniformity of illumination, acoustic frequency mechanical stress (1~8 KHz) is applied to the delivery multimode fiber to modulate its index of refraction and thus the mode patterns. The comparison of the image quality before and after speckle reduction is illustrated in Fig. 2.

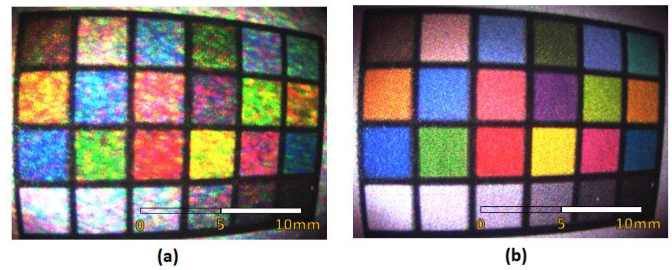

Fig. 2. Comparison of speckle reduction (a) without and (b) with acoustic frequency mechanical stress modulation on the multimode delivery fiber. To better illustrate the speckle effects, only the images from the rigid scope are shown.

The images were acquired by using a 10K pixels high definition image guide (HDIG) of focal distance 5 mm and a rigid scope of 2.7 mm diameter with a focal distance of 30 mm, each having a viewing angle of ~65°. In this work, we choose to use a monochrome CMOS camera (FL3-U3-13Y3M-C, FLIR Integrated Imaging Solutions Inc., Canada) placed at the back end of the HDIG instead of a traditionally used color CMOS camera in order to achieve better sensitivity and higher spatial resolution. Notably, for a color CMOS sensor, a color filter mosaic of tiny color filters is placed over the pixel sensors to capture color information as shown in Figure 3(a). The demosaicing algorithm is then used to create (interpolate) data that is not actually captured to reconstruct the color image. However, loss of photons, because of filtering, affects the sensitivity and quantum efficiency (QE) of the photosensors. There could be some noticeable effects or artifacts if the color information in the scene changes rapidly (specifically, near the resolution limit of the sensor), since the spatial frequency of (typically) red- and blue-filtered sensor pixels is only 1/4 the sensor's absolute spatial frequency. As a result, the color-specific spatial resolution of the output image is correspondingly reduced.

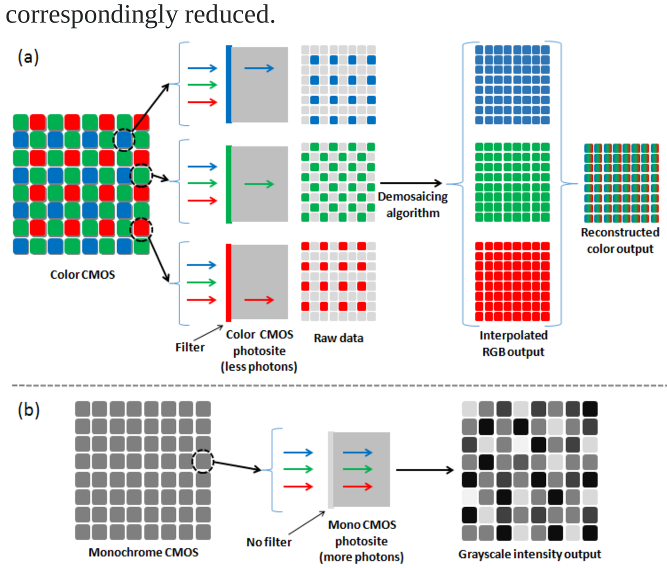

Fig. 3. Principle of (a) color and (b) monochrome CMOS camera

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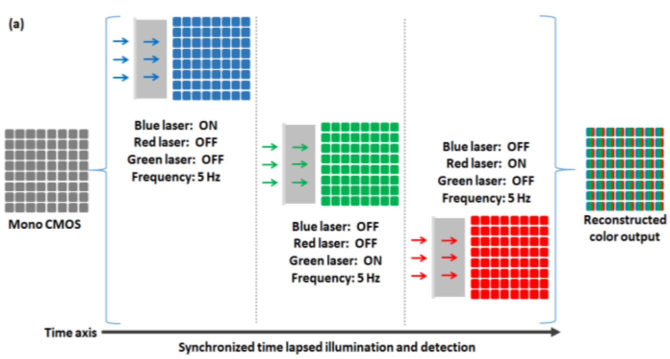
In contrast to a color CMOS camera, a monochrome CMOS camera avoids the color filter arrays thereby increasing the sensitivity even at much lower illumination conditions by allowing more photons to reach the photosensitive area as shown in Fig. 3(b). Furthermore, it is unnecessary to use the demosaicing algorithm in case of a monochrome CMOS camera for final image reconstruction. Instead, the intensity recorded in each of the photosensitive area represents the values at each pixel, which can be subsequently used for a grayscale image reconstruction, providing better resolution than a color one under the same construction geometry.

Since a monochrome CMOS camera can provide only grayscale images, in this work a high frame-rate monochrome CMOS camera is used in combination with synchronized and spectrally resolved illumination scheme in order to reconstruct a color image or video as depicted in Fig. 4(a).

In this scheme, the sample is illuminated sequentially with the three colors of the RGB lasers and the gray scale intensity information corresponding to the specific color illumination sequence is recorded. Each of the grayscale images is then color-coded accordingly. Notably, the narrow band spectrum of the RGB laser acts as a very important illumination characteristic that gives accurate information for color-coding. The whole process is controlled by a dedicated software and by time lapsed illumination; color recording is achieved through combining every three consecutive Red, Green, and Blue images to reconstruct one full color frame as shown in Fig. 4(b). In addition, the software is also capable of controlling the duty cycle of all the three laser wavelengths individually for better adjustment of both the exposure and white balance. It is noteworthy to mention that the synchronized illumination and image acquisition allows for the reconstruction of a color image only from that region of interest, which is illuminated by the time-lapsed illumination scheme. An example is shown in Fig. 4(c).



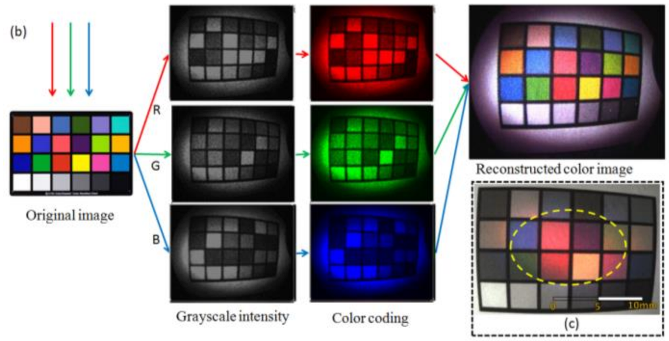


Fig. 4. Showing the temporal sequence in the synchronized illumination and the color allocation. (a) Synchronized time lapsed illumination and detection scheme adopted in this work. (b) The sample (color checker) is illuminated in the sequence of blue, green and red at the rate of 5 Hz each. The corresponding grayscale images of the sample is recorded and subsequently color-coded and the original sample image is reconstructed by a dedicated software. (c) Color output (image) is reconstructed only for the circled area that is synchronously illuminated by the RGB laser. Note that outside of the circled area, color information is lost, as expected.

III. RESULTS AND DISCUSSION

The illumination performance such as color rendering index (CRI), emission spectra, correlated color temperature (CCT), chromaticity coordinates and illuminance [22], [23], [24], [25], [26], of a reference white LED light source and the RGB laser are evaluated with the chromameter (SRI2000, Optimum Optoelectronics, Taiwan). Notably, CRI reflects the quality of an illumination system and measures the capability of reproducing the colors of an object faithfully or realistically when illuminated by that light source as compared to when illuminated by a reference source of comparable color temperatures [22], [23]. However, this conventional definition of CRI is used to compare the color deviations between continuous spectrum light sources. In case of discontinuous spectrum, e.g., for a multi-wavelength laser system, its ability to reveal the natural color of an object can be evaluated by comparing the gamut area (GA) [17], [27] with that of the CIE1931 color space [28], [29]. In order to measure CRI of the RGB laser system, a CIE1931 chromaticity diagram with grid lines as shown in Fig. 5 is utilized to find the area surrounded by the triangle, the vertices of which is decided by the wavelengths of the three lasers. The CRI of the three-wavelength laser system is obtained by calculating the ratio of the area of the triangle (i.e. the GA of the RGB laser) and that of the CIE1931 chromaticity diagram.

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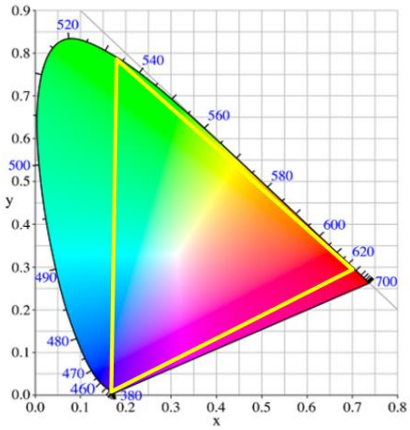

Fig. 5. The gamut of the RGB laser system [17]

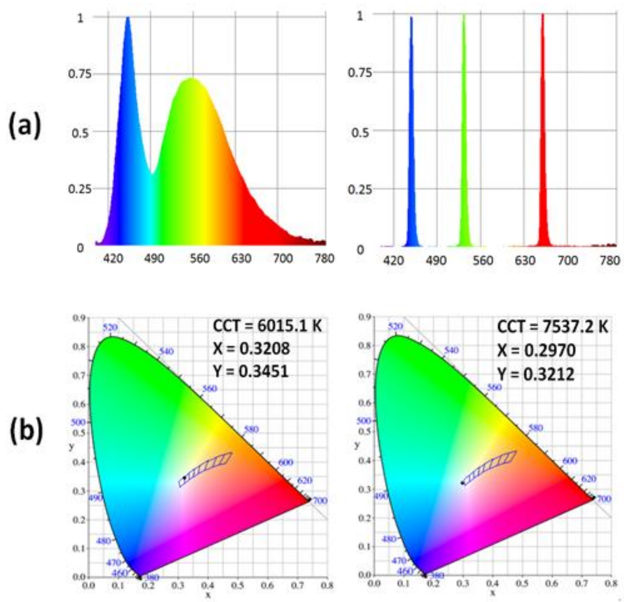

Fig. 6. (a) The emission spectra of the white LED (left) and the RGB laser (right). (b) The CIE1931 chromaticity coordinates of the white LED (left) and the RGB laser (right). Notably, the chromaticity coordinates and CCT of standard D65 light source are (0.3127, 0.3290) and 6500K respectively.

Fig. 6 compares the color co-ordinates and CCT between a medical grade white LED and the RGB laser on a CIE1931 chromaticity diagram in terms of emission spectra (a) and quality evaluation (b). After tuning, the color coordinate of the RGB laser system is found close to that of the standard light source D65 [30]. On the contrary, the color coordinate of the white LED is determined by the ingredients of phosphor. While the illuminance of the RGB laser source and LED source was found to be comparable in free space, the illuminance of the LED source dropped to a very low level due to the lack of coherence and thus the huge coupling loss, which is not the case for the RGB laser source. The measured data is listed in Table 1.

Next, to evaluate the imaging performance, a comparative analysis was conducted between mono CMOS with the RGB laser illumination and color CMOS with the LED illumination by using the Macbeth color checker [31] of size 18x12 mm at comparable illumination conditions. As shown in Fig. 7, the

TABLE I
COMPARISON OF THE ILLUMINATION AND COLOR PERFORMANCE OF MEDICAL GRADE WHITE LED AND RGB LASER SYSTEM

Light Sources	RGB laser	White LED
Maximum illuminance at a distance of 5 mm from the source without coupling to the fiber probe (lux)	61000	52300
Maximum illuminance at a distance 5 mm away from the fiber probe (lux)	18290	50
CRI (without fiber probe)	74	75.8
CCT (K)	7537.2	6015.1

RGB laser module together with the monochrome CMOS camera provides better color reproducibility and saturation (RMS difference with the original RGB value = 38.65) than white LED with color CMOS camera (RMS difference with the original RGB value = 50.95).

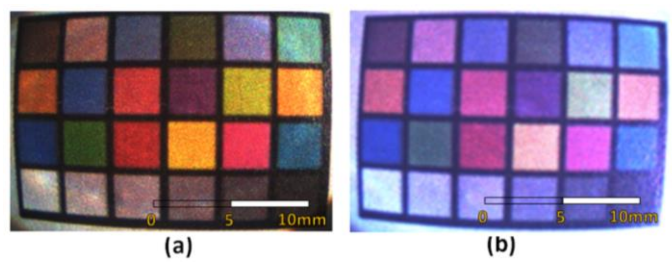

Fig. 7. Comparison of images for (a) RGB laser with mono CMOS camera versus (b) standard white LED with color CMOS camera. Note that lower color saturation may be attributed to the color filter on the color CMOS camera.

To further emulate the surgical scenario, a 4-month old male nude mouse was used as the specimen, indicated as SM in Fig. 1. The illumination conditions for the *in vivo* experiment were kept identical to those used for the color checker. Notably, the animal experiment was reviewed and approved by the Institutional Animal Care and Committee of National Yang-Ming University. In order to achieve a bigger field of view and observe organs of larger size, the 2.7 mm rigid endoscope was used for this purpose and placed approximately 30 mm away from the targeted animal. After the experiment, the body of the mouse was checked and found that the illumination from the RGB laser did not cause any detectable effects on the tissues or organs.

The corresponding imaging using animal model is given in Fig. 8. The results clearly demonstrates the improvement of image rendering quality of mono CMOS with the RGB laser illumination in Fig. 8(b) as compared to that of color CMOS with LED illumination in Fig. 8(a). Further, illumination using a white LED produced more glare as compared to RGB based illumination which may obscure the distinction among various tissues. Most importantly, the sequenced illumination and synchronized detection scheme presented here has the potential for narrow band imaging as shown in Fig. 8(c). The observed grainy effect in Fig. 8(c) is attributed to lower illumination used and the threshold parameters chosen for the imaging processing software. Notably, as a technology for optical image contrast enhancement, NBI exploits the property of hemoglobin in the blood vessels to strongly absorb blue and green wavelengths, to further improve the visibility of the capillary blood vessels in the surface of mucosa [32]. In contrast to the conventional NBI imaging techniques that use optical filters to generate NBI light by eliminating all wavelengths except blue and green, the scheme presented here uses narrow band laser sources allowing to readily achieve NBI illumination conditions.

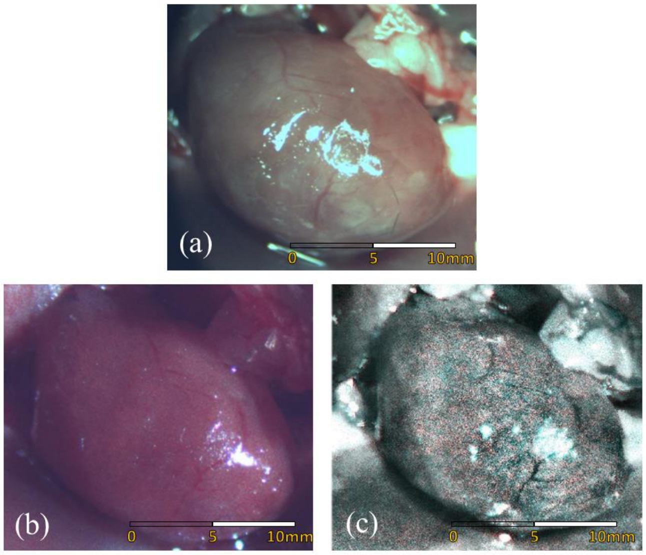

Fig. 8. The nude mice' heart imaged by (a) color CMOS with LED illumination, (b) mono CMOS with RGB laser illumination and (c) image contrast of (b) is enhanced through narrow band imaging (NBI). The illuminance from the RGB laser and the LED is kept the same during testing. The 2.7 mm rigid endoscope is used for collecting the images in this case.

Our results show that the narrow band RGB laser module can be used as a high performance light source capable of more efficient light coupling due to its spatial coherence as compared to conventional LED and Xenon lamp sources. Moreover, the structure of the RGB laser module is far more compact and robust than that of the supercontinuum source used in a previous study [19]. Importantly, the use of the RGB laser in this work not only proved to be a cost effective and compact alternative light source for generating white light by multiplexing the red, green and blue laser wavelengths, but also paved the way for downsizing the overall diameter of the endoscope to an ultrathin scale of approximately 0.49 mm. We have addressed the problems of speckle and small divergence angle from the RGB laser module by using mechanical modulation at acoustic frequency and a custom designed collimator, respectively, to provide suitable illumination for the ultrathin endoscope. Furthermore, spectrally resolved high-resolution images were obtained by synchronizing illumination with mono CMOS cameras. In addition, color-coding in spectral imaging can be paired to interested biological tissues to optimize contrast information. For instance, we have demonstrated the applicability of our endoscope for narrow band imaging (NBI) [32], [33], [34], which is another striking feature that makes it an excellent tool for high contrast imaging of blood vessels in mucosa.

IV. Conclusion

The ever-decreasing size of the CMOS imaging sensors always demanded for the innovation in reducing the size of the light guide so that the footprint of the illumination components would not restrict an endoscope in exploiting the advantages of such ultra-compact CMOS sensors. In this work, we implemented all fiber based synchronized illumination and image acquisition to reduce the overall diameter of the endoscope to 0.49 mm. Such a combination would enable the user to place the ultraslim endoscope along with other medical tools to explore the interior part of some organ through narrower pathway or narrower orifice [17].

In summary, with the improved beam divergence angle and the uniformity of the illumination intensity obtained by using a custom built collimator and acoustic vibration scheme, the RGB laser module presented as a compact and efficient light source is ideally compatible to the miniaturized CMOS image sensor to form an ultra-slim endoscopic system for diagnosis and minimally invasive surgery. Besides, the portability and the lower price of the RGB laser module will make it more competitive than other commercial products. We anticipate that these unique advantages of the all fiber based ultraslim endoscope equipped with coherent narrow band RGB laser for illumination and mono CMOS for imaging will provide unprecedented opportunities in medical applications.

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the optical element and is expected to make the optical and medical fields more integrated. It also focus on laser light source which provide optional wavelength to synchronize digital signal for spectral images. Mr. Zhan-Yu Chen is also a recipient of the License of Radiology Technologist, Taiwan.

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Miniaturization of instruments is one of the main developments of modern surgery, especial in minimal invasive surgery that includes many categories. Recently, he cooperates with the laboratory of Prof. Fu-Jen Kao to improve the effectiveness of instrument in minimal invasive surgery, focused on the endoscope. We tried to minimize the size of light source of endoscope during the minimal invasive surgery using different methods and got some initial experiences.

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