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Endoscopic hyperspectral imaging: light guide optimization for spectral light source

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ABSTRACT

Hyperspectral imaging (HSI) is a technology used in remote sensing, food processing and documentation recovery. Recently, this approach has been applied in the medical field to spectrally interrogate regions of interest within respective substrates. In spectral imaging, a two (spatial) dimensional image is collected, at many different (spectral) wavelengths, to sample spectral signatures from different regions and/or components within a sample. Here, we report on the use of hyperspectral imaging for endoscopic applications. Colorectal cancer is the 3rd leading cancer for incidences and deaths in the US. One factor of severity is the miss rate of precancerous/flat lesions (~65% accuracy). Integrating HSI into colonoscopy procedures could minimize misdiagnosis and unnecessary resections.

We have previously reported a working prototype light source with 16 high-powered light emitting diodes (LEDs) capable of high speed cycling and imaging. In recent testing, we have found our current prototype is limited by transmission loss (~99%) through the multi-furcated solid light guide (lightpipe) and the desired framerate (20-30 fps) could not be achieved. Here, we report on a series of experimental and modeling studies to better optimize the lightpipe and the spectral endoscopy system as a whole. The lightpipe was experimentally evaluated using an integrating sphere and spectrometer (Ocean Optics). Modeling the lightpipe was performed using Monte Carlo optical ray tracing in TracePro (Lambda Research Corp.). Results of these optimization studies will aid in manufacturing a revised prototype with the newly designed light guide and increased sensitivity. Once the desired optical output (5-10 mW) is achieved then the HIS endoscope system will be able to be implemented without adding onto the procedure time.

Keywords: Hyperspectral, Fluorescence, Colorectal, Cancer, Spectral Imaging, Endoscopy, Endoscope, Spectroscopy

1. INTRODUCTION

1.1 Hyperspectral Imaging Biomedical Applications

Hyperspectral imaging (HSI) produces a spectral image cube from an image acquired over multiple wavelengths. Each pixel of the overlapped image has a specific spectral signature. The substrates of the sample often have unique reflectance and fluorescence spectral signatures. This technique has been used in remote sensing satellite imagery to determine biological differences such as vegetation or bodies of water. The same application is used in archaeology to map ancient architecture[1,2] and in food processing to estimate food spoilage or caloric content (fat, sugar).[3] Other applications include historical documentation and art preservation through enhancing worn images and subtracting wavelengths to visually diminish staining.[4,5] A further area of interest for hyperspectral imaging is biomedical uses such as microscopy and endoscopy which will be the focus of this document.

The work done in our research group integrates hyperspectral imaging into microscopy and endoscopy. A desired outcome for biomedical imaging is collecting live cell and real time data. The non-biomedical applications described in the preceding paragraph are not time dependent and hence acquisition-time related signal-to-noise characteristics are not as critical as they are in biomedical microscopy and endoscopy. Previous work in our lab has demonstrated that biomarkers can be detected with high fidelity using HSI.[6–8] Historically, biomedical hyperspectral imaging has been implemented using emission scanning where the substrate is excited at one wavelength and the fluorescence emission spectrum is detected. Here, we use a new technique developed by our lab called excitation scanning, where the substrate is excited in a wavelength-dependent manner across a range of many excitation wavelengths and the entire emission signal is detected; this is repeated for every excitation wavelength. [9]

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Our initial prototype to demonstrate the feasibility for excitation scanning HSI on a microscope system used for the excitation scanning technique used a xenon arc lamp and an array of thin film tunable filters to provide the correct wavelength and bandpass for excitation data collection.[9] The center wavelength of the band was tuned using changes through a mechanical motor switching angles and positions of the thin film tunable filters.[10,11] The time dependent factor in the previous spectral light source was the mechanical switching of the motor. The requirements for a 16 wavelength unit producing 20 image cubes per second would be 320 wavelength frames per second. Unfortunately, there is not a current filter switching and filter rotating motor combination rated for this speed. Hence, we have focused on designing and testing of a spectral light source that will meet the speed requirements for real time hyperspectral imaging in endoscopy.

1.2 Previous Prototype for Spectral Endoscopy

To meet the requirements for high-speed wavelength switching a prototype endoscopic spectral light source was designed with 16 wavelength specific LEDs in place of the xenon arc lamp and thin film tunable filter array. This removes wavelength switching delays associated with the mechanical motor and instead uses electronically triggered LEDs to reduce acquisition times of each wavelength. Combining LED outputs was accomplished with a branched solid light guide. Every LED has a respective branch which all connects to a common output that is coupled with the endoscope's fiber optic input. The LEDs and solid light guide components were added to reduce acquisition time for a spectral image cube.

LED incorporation required supplementary electronic constituents to provide analog and digital inputs, constant driven current to the LEDs and the LED array. The analog and digital inputs were traced from the computer to the respective LED current driver (RCD-24-0.70, RECOM) on the current driver board designed and printed using printed circuit board (PCB) software (Pad2Pad, Pad2Pad Inc.). The current from each current driver was traced to the corresponding LED on the array board (Pad2Pad). The PCBs and computer cards were connected using D-SUB connector wires to associate a specific pin/wire combination to its respective LED. The solid light guide was bracketed to the electronics using 3D printed CAD designs (Inventor, Autodesk). A branched alignment bracket attached the LED array to the branched end of the light guide and a common output plate attached the light guide to the coupling mechanisms of a commercially built endoscopic light source (CLK-4, Olympus). The added features were all housed in a rack-mountable enclosure to protect from biological exposure during the testing phase.

The resulting spectral light source prototype was tested for correct electronic controls and intensity transmission for imaging.[12,13] The electronic tracing during construction of the device allows for software inputs and controls to trigger the LEDs and camera acquisition. The electronic triggering of the LEDs allowed for microsecond intervals of wavelength cycling correlating to millisecond exposure times and image acquisition. The transmission throughput for LED intensity was measured using a spectrometer (QE65000, Ocean Optics) and integrating sphere (4P-GPS-030-SF, Labsphere) calibrated to a NIST-traceable response using a standardized light source (LS-1-CAL, Ocean Optics). The resulting intensity indicated a transmission loss of 99% through the light guide and endoscope. The reasons for high transmission loss are the viewing angles of the LEDs and the lack of total internal reflection (TIR) of light through the light guide. The input of each branch does not collect the entire viewing angle of the LED's lens, losing light at the start. Due to the curvature and intersection points of the solid light guide there are various areas for refraction and transmission loss. In the following paper, we describe a process for optimizing the light guide to diminish transmission loss through the device to allow for excitation scanning hyperspectral imaging. Additionally, the work shown below will provide image results of the device before optimization to show the capabilities already available.

2. METHODS

2.1 Optical Transmission Modeling

The first stage of optimization was to model the solid light guide using raytracing software (TracePro, Lambda Research). TracePro uses Monte Carlo analysis to determine the amount of light measured versus the amount of light emitted. The light guide was designed in Autodesk Inventor and imported to TracePro as a .sat file for simulations. The LEDs used in the prototype were replicated using the source property option. The input of the endoscope was made as the interrogation surface to measure the amount of light transmitted.

A baseline model was defined for a simulation that closely resembles the measured intensity of the current device (Figure 1). The spectral irradiance properties of a 405 nm LED (SMB1N-405V-02, Roithner LaserTechnik) were configured in TracePro as the emitter. The LED was positioned 5 mm away from the branch's input. The interrogation surface was created as a cylinder with a diameter of 5 mm and positioned 3 mm away from the light guide's output to simulate the endoscope optic input. The interrogation surface was marked as an exit surface and a perfect absorber. The raytrace transmission measurement was averaged over 100,000 rays run.

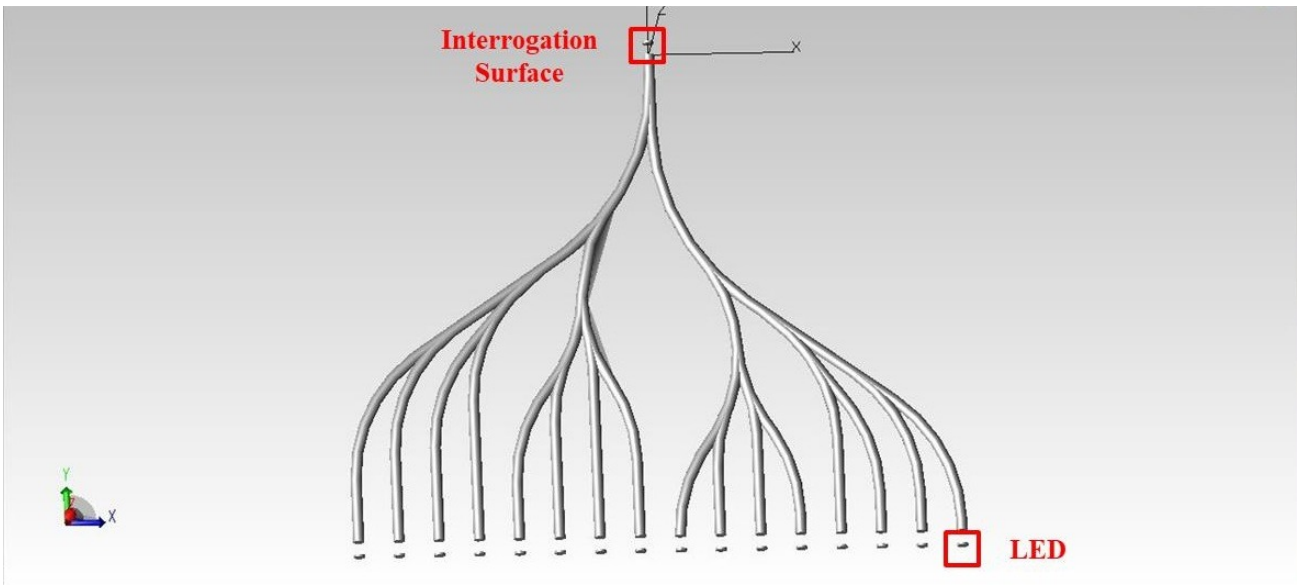

Figure 1: The TracePro window displaying the baseline model of the solid light guide and highlighting the LED and interrogation surface. This view is prior to raytracing

As discussed above, most LEDs have a dispersive emission angle greater than the acceptance angle of the solid light guide branch and this accounted for much of the light loss. To attempt to collect more of the emitted light, a lens was added in front of the LED to collimate the light inputted to the light guide branch. The lens (69858 Aspheric Lens, Edmund Optics) was placed 9 mm away from the branch and the LED was placed 9 mm away from the lens.

2.2 Emission-Scanning Tissues

While the software optimization is still ongoing, we also performed initial feasibility testing of the initial prototype HSI endoscope. A section of descending and spiral colon was obtained from an unrelated study at the University of South Alabama College of Medicine Vivarium. The sample was rinsed thoroughly with PBS and sutured at one end. Reflectance was imaged using 420, 470, 525, 670 and 850 nm wavelengths at 0.3 mW. An image was captured using acquisition speeds of 80 and 100 ms. The intensity was increased to 0.4 mW using the 525, 670 and 850 nm wavelengths and imaged at speeds of 50 and 100 ms. The excitation wavelengths used for emission images were 420, 470 and 525 nm at 0.3 mW. The images were collected using acquisition times of 1 and 10 s. A 560 nm long-pass filter was placed before the camera to capture emission.

3. RESULTS

3.1 Light Guide Modeling Performance

The baseline model of the solid light guide using TracePro resulted in 3.87% light transmitted to the exit (common) branch of the light guide (Figure 2). This is compared to the experimental 2.1% transmission measured using the spectrometer and integrating sphere. The additional losses are likely due to non-ideal performance of the manufactured light guide.

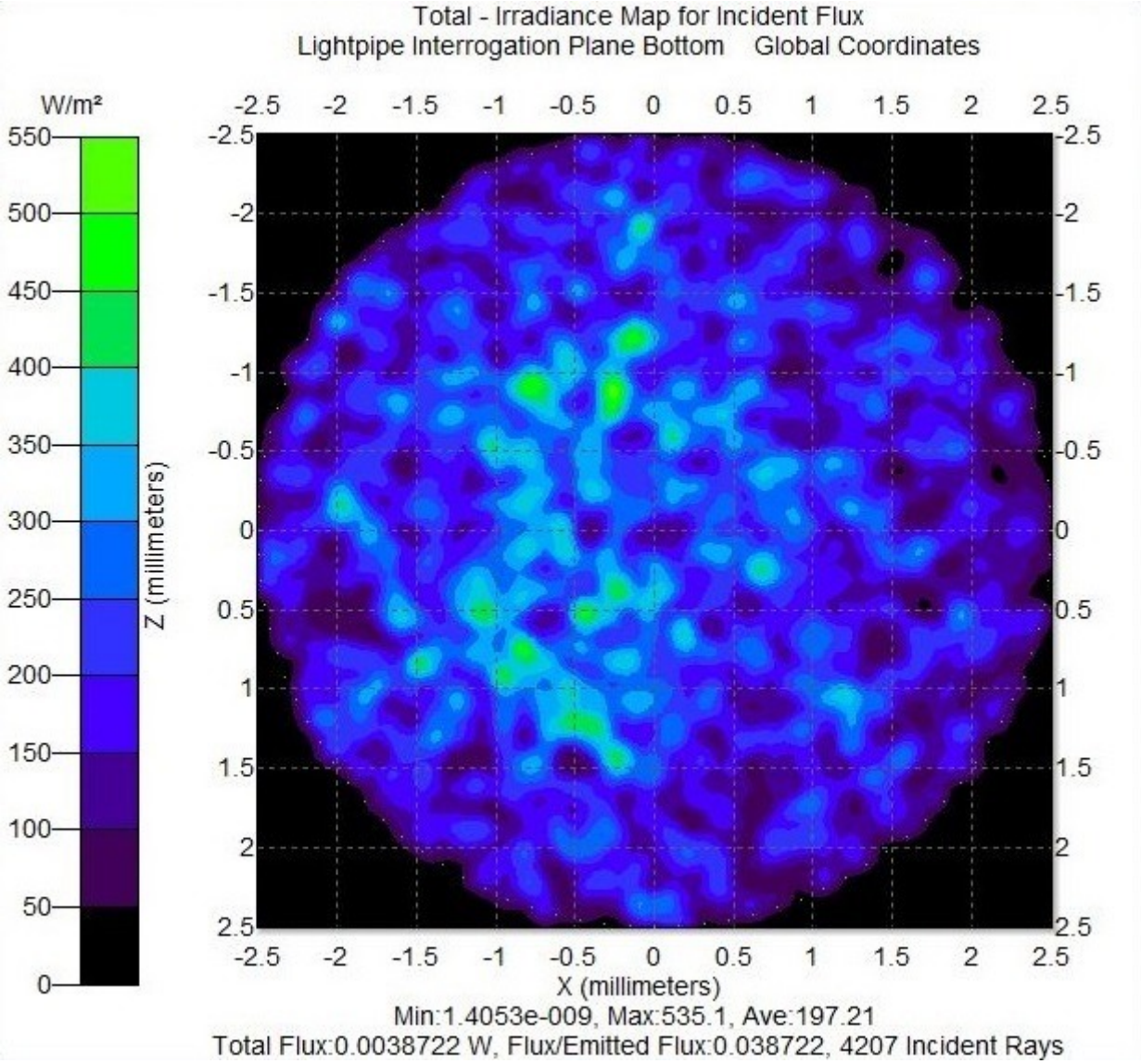

Figure 2: The irradiance map for the baseline lightpipe design TracePro. The flux/emitted flux indicates transmission of light to the interrogation surface (to resemble the endoscope input).

The addition of a lens resulted in a 6.59% transmission which is almost double the baseline measurement (Figure 3). The design specification for the solid light guide is 10% transmission. This specification is based on achieving an illumination intensity of 10 mW at any of the wavelengths and the fact that LEDs are available with output powers ranging from 100 mW to 1 W. Therefore, adding a coupling lens to the light path could achieve half of the intended specification.

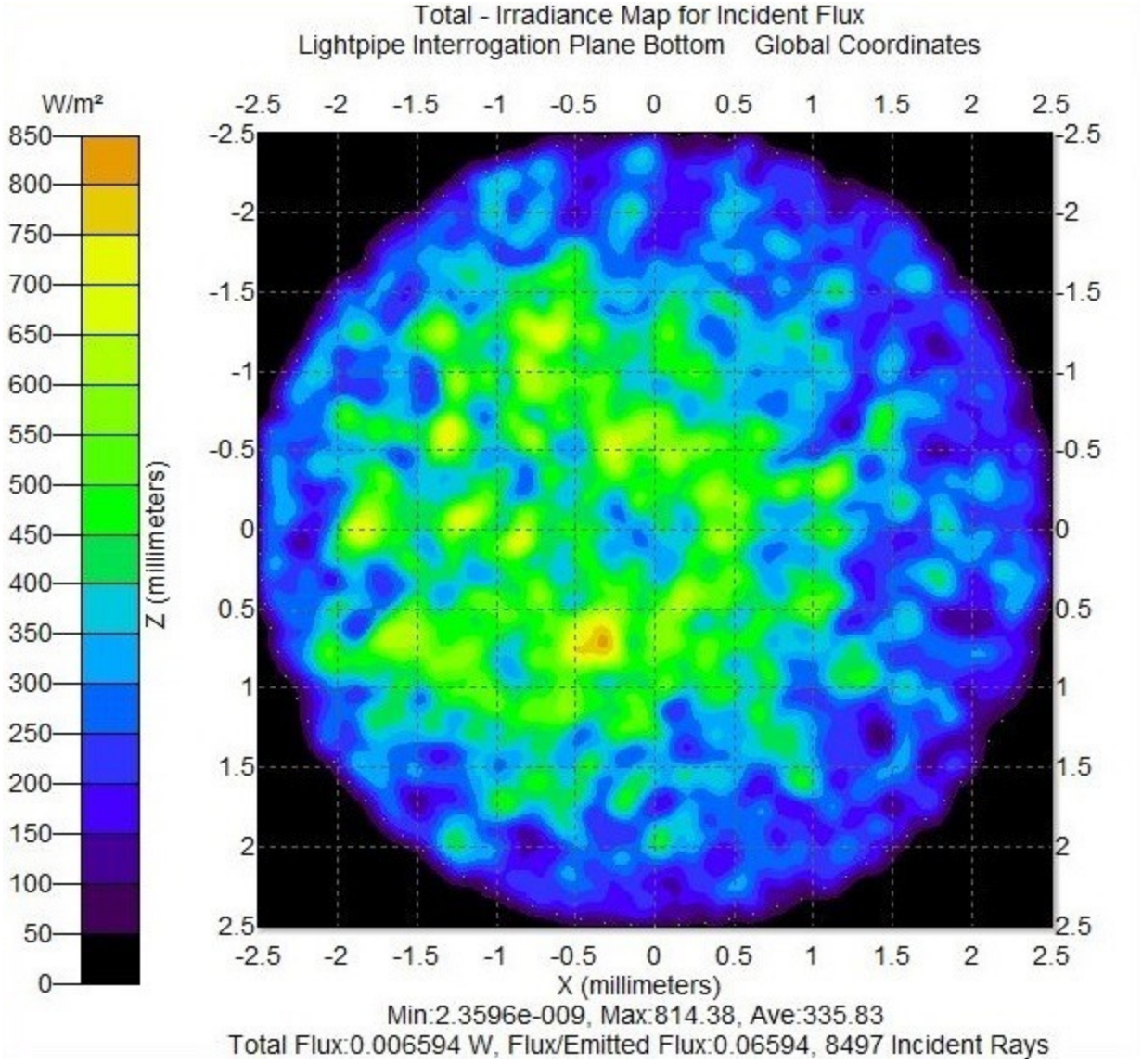

Figure 3: The irradiance map for the TracePro model of the solid light guide including a lens to collimate the LED light into the light guide.

3.2 Emission-Scanning Hyperspectral Images

Feasibility testing of the initial spectral endoscopic light source prototype demonstrated reflectance imaging (Figure 4) and emission imaging (Figure 5). The tissue reflects the flesh coloration and folding of colorectum walls during imaging. The pixilation is a result of the limited light captured per cycle hence, another reason why the optimization is important at this time.

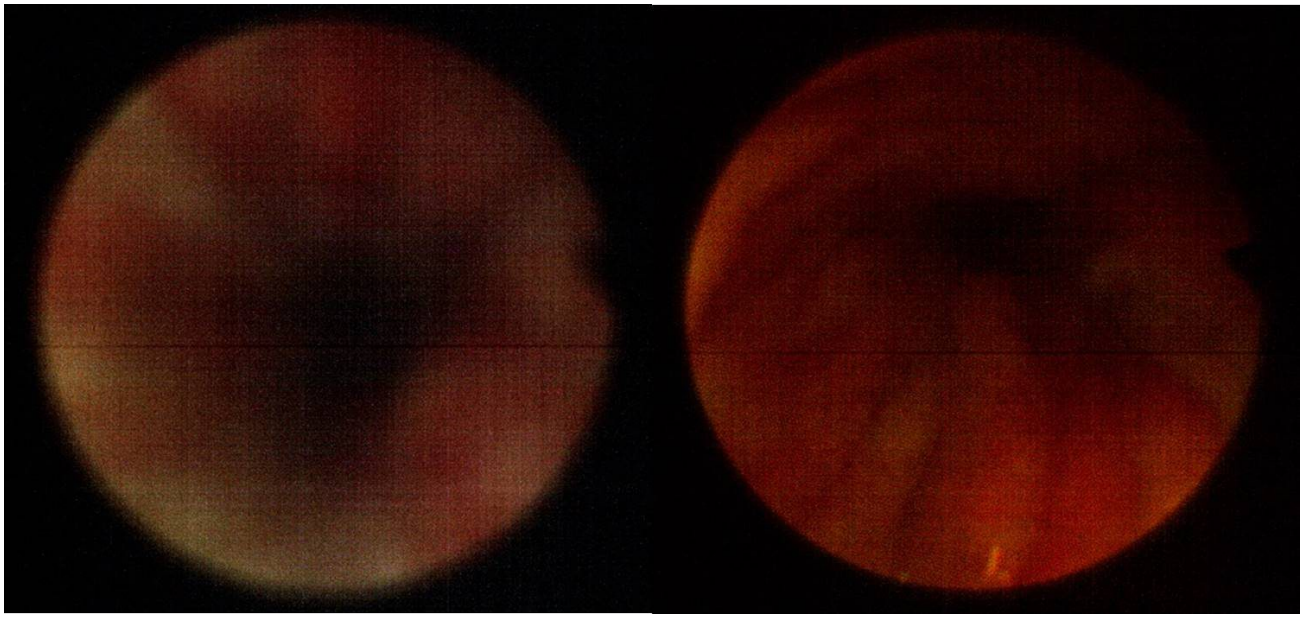

Figure 4: Reflectance images of pig colon at 0.3 mW and an acquisition time 100 ms (Left), and at 0.4 mW and an acquisition time 50 ms (Right).

The emission image at 1 second acquisition produced a poor signal-to-noise ratio, so an image at 10 second acquisition was also acquired (Figure 5). A split image is shown to provide a view of each individual wavelength image. Once again, the acquisition time for this initial test was too long for real time imaging in endoscopy and in order to increase the framerate, the intensity output needs to increase through software optimization.

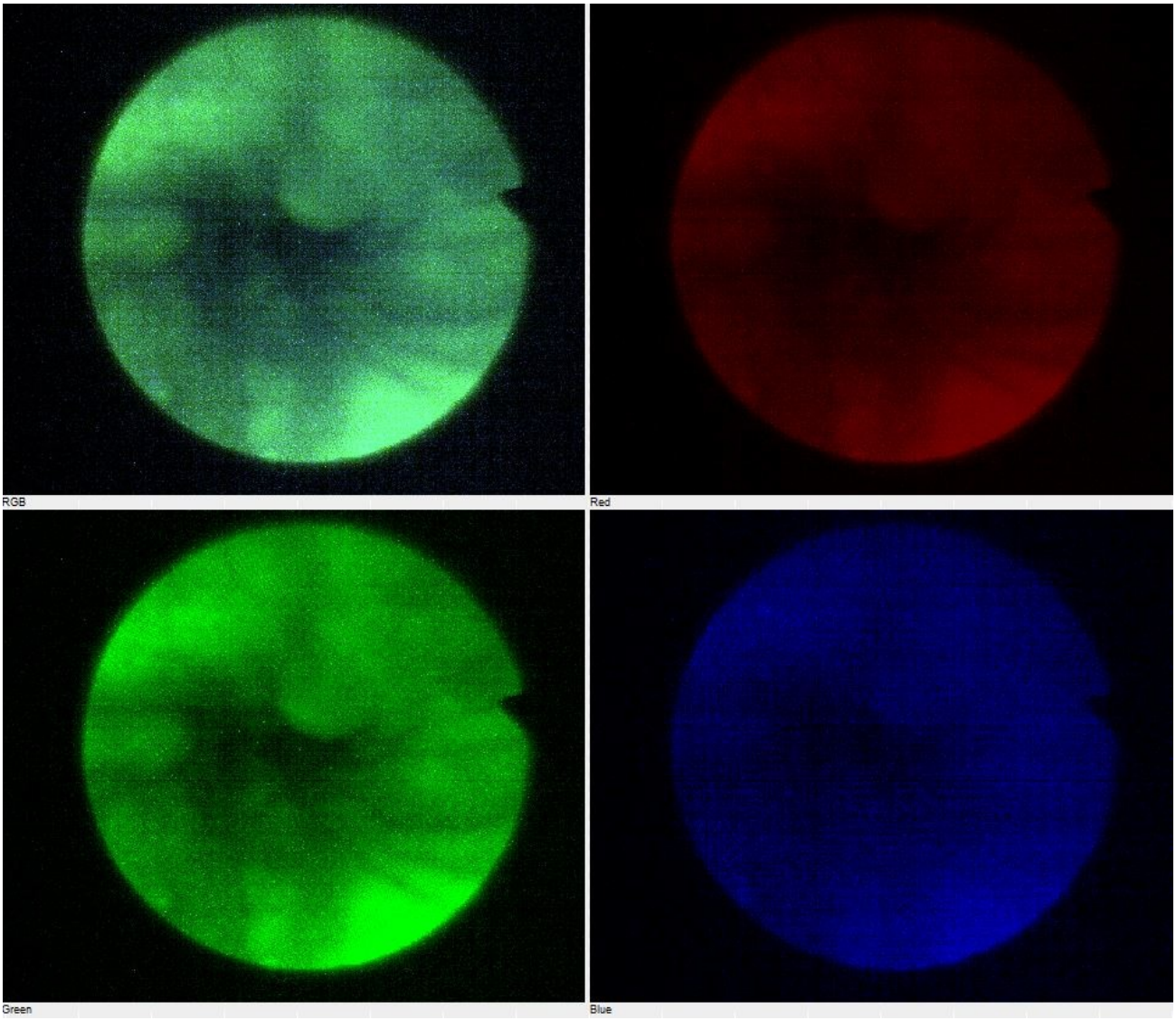

Figure 5: False-colored display window of the 10 second emission spectral image with the super-imposed false-colored image (top/left) and the individual wavelengths in the other windows.

4. DISCUSSION

The result from the images captured at low intensities (~0.4 mW) proves the spectral light source is capable of hyperspectral imaging in endoscopy. Once the transmission loss is minimized, a fully developed excitation scanning test will produce promising results. One feature of using LEDs, not discussed previously, is the ability to normalize intensities prior to imaging. The analog input for each current driver allows the user to set a maximum intensity output for imaging. Previous work has required a background image and normalization after testing. This feature is another way to decrease the amount of time it takes to acquire an image and providing the data to the user during a procedure. The testing for excitation scanning hyperspectral imaging moves us towards the overall goal of determining spectral differences in normal and cancerous tissues in endoscopy during a procedure.

In order to reach the goal of a minimum transmission of 10% we will investigate the geometry of the solid light guide. The focus will be on the degree of each bend and the reflection of each intersection point. The goal of the geometry changes will be to maintain TIR until light exits the common branch output. The higher transmission simulated will

move the new design into production to be tested in the spectral device. Depending on the geometric changes, the LED array board and the alignment brackets will have to be revised to accommodate the new light guide. The established light guide will then be tested for intensity output to compare to the simulated results. Successful results will allow for some LED input adjustments and continuation of testing the excitation scanning technique.

After optimization, a comparison test will be produced to test hyperspectral imaging against white light endoscopy, narrow band imaging, autofluorescence imaging and virtual chromoendoscopy. These tests will measure the specificity, sensitivity and accuracy of each modality within the device. Specified LEDs will be combined to replicate each modality so the same spectral light source will produce each one with a software setting adjustment. This work will move us into the clinical trial stage to collect substantial data providing feedback that hyperspectral imaging will increase detection accuracy of abnormal growth within endoscopy.

5. ACKNOWLEDGEMENTS

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