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Hyperspectral Imaging for In Vivo Detection of Trachea Tissue

William Y. Christensen
Department of Electrical Engineering and Computer Science
United States Military Academy
West Point, New York 10997

Faculty Advisor: Dr. Brian Souhan

Abstract

The purpose of this project was to build a device that could detect tracheal and esophageal tissues to assist in tracheal intubation procedures. The devices built focused on filtering specific spectra of reflected light. The first device projected white light from LEDs onto the tissue samples with the reflected light being measured by photodiodes after passing through optical filters. The second device projected white light from LEDs and captured and transmitted the reflected light via optical fiber to photodiodes with similar optical filters mounted on them. Although a delay in receiving the tissue caused the results for the first device to be inconclusive, the second device detected trachea tissue with 100% accuracy.

Keywords: Tracheal intubation, Reflection Spectrometry, Hyperspectral Imaging

1. Introduction

Tracheal intubation involves inserting a tube into the trachea to allow medical procedures such as general anesthesia and ventilation. The process of intubation requires proper insertion into the trachea. However, incorrect insertion into the esophagus occurs up to 23% of all intubation attempts [1]. Devices that can detect esophageal and tracheal tissue could allow a much higher success rate of intubation. Previous research utilizing hyperspectral imaging has shown noticeable differences in the reflectance spectra of esophageal and tracheal tissue at three distinct wavelengths—543 nm, 561 nm, and 578 nm. A prototype utilizing discrete components examining these three wavelengths successfully detected tracheal tissue in 18 out of 19 trials [2]. Despite the positive results, several issues needed to be addressed, including components being too large for practical use, poor intensity returns when using optical fibers, and the probes not pointed perpendicular to the tracheal wall.

The purpose of this project is to build a sensor that can accurately distinguish between tracheal and esophageal tissues by fabricating and testing new prototypes that point the probe perpendicularly to the tracheal wall to maximize reflectance return. Filtered light will activate photodiodes to generate current allowing a simple logical comparator to determine the tissue type. Despite several trials, the first device using discrete components failed to distinguish the two different tissues. The secondary design utilizing an optical fiber showed more promising results. Using a ratio test of the three different wavelengths of light, the new device detected the trachea tissue 100% of the time. However, during these tests, only tracheas were examined as esophagi were not available for testing.

The key aspect of this project is the composition difference between esophageal and tracheal tissues. Their unique compositions give each a distinct reflectance spectrum. The specific difference comes at the noticeable peaks and drops of the two different tissues' reflectance of light at the wavelengths of 543 nm, 561 nm, and 578 nm. Figure 2 illustrates these peaks and drops. The fundamental concept of this project focuses on leveraging these differences. Intubation processes rely solely on the skill of the medical professional for proper insertion of tubes into the trachea. A device that could detect the difference between the two would assist in the intubation process.

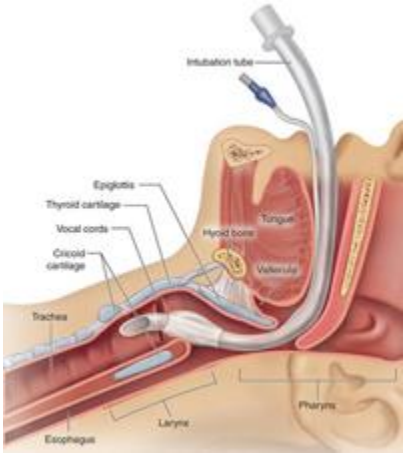

Figure 1: A cross sectional view of an intubation process [3].

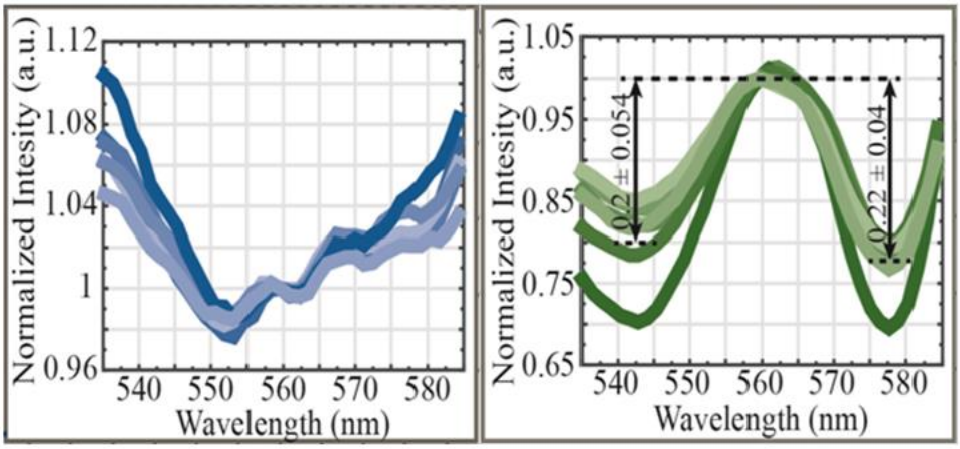

Figure 2: The graph on the left represents the reflectance spectrum of esophageal tissue; the graph on the right shows the reflectance spectrum of tracheal tissue [4].

The esophageal tissue (depicted in blue) has a noticeable dip in normalized intensity between 550 nm and 570 nm unlike the tracheal tissue (depicted in green) which exhibits a peak in between those wavelengths. Additionally, there exist local minima in the normalized reflectance intensity of the tracheal tissue near 540 nm and 580 nm unlike the normalized reflectance intensities of the esophageal tissue that rises at those specific wavelengths.

By using filters to isolate certain spectrums of light, it is possible to use photodiodes and digital logic to distinguish between these two tissues [2]. The figure below shows the bandwidth and transmission of the 561 nm filter (LL02 561) obtained from Semrock and used in the secondary design.

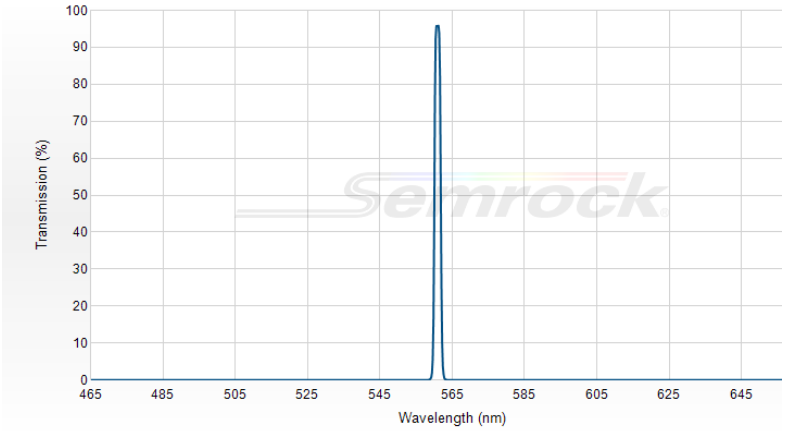

Figure 3: 561 nm filter bandwidth and transmission [5].

Earlier designs sought to utilize the three large differences between the normalized intensities near 543 nm, 561 nm, and 578 nm (see figure 3) to distinguish between the tissues. However, in order to reduce overall size, the first design presented here only utilizes the differences at 543 nm and 561 nm since the marginal utility of comparing the three bands is minimal to comparing the 543 nm and 561 nm bands [2]. With the second design again moving the filters and photodiodes outside the tube, the need to minimize size was less, and therefore all three bands were used for those trials.

2. Probe Design

The two critical aspects of the first design were size and maximizing the photodiode output. The maximum diameter of the trachea is typically 21 mm [7]. This constraint limits the size of discrete components that can be placed on the device. Thus, the device must house most of the circuitry outside of the tracheal tube.

A gain circuit using a non-inverting amplifier design with the LT1226 operational amplifier was designed to act as an amplifier for the photodiode currents. This is a high gain and low noise device used to maximize the small signals generated from the reflectance of the tissue.

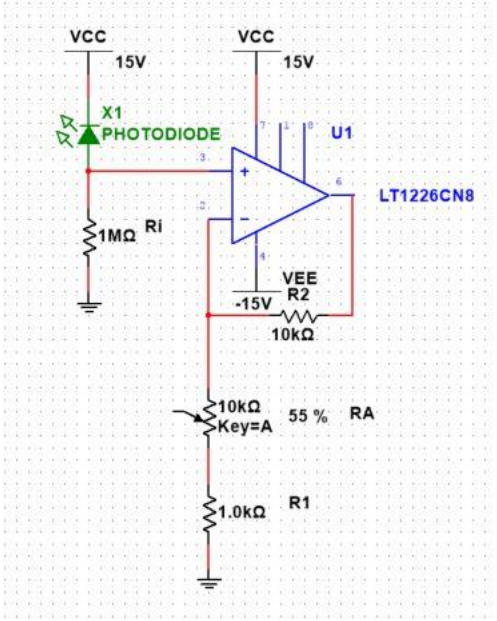

Figure 4: High-gain, low-noise circuit for detecting the two signals from the photodiodes.

2.1. Discrete Probe Design

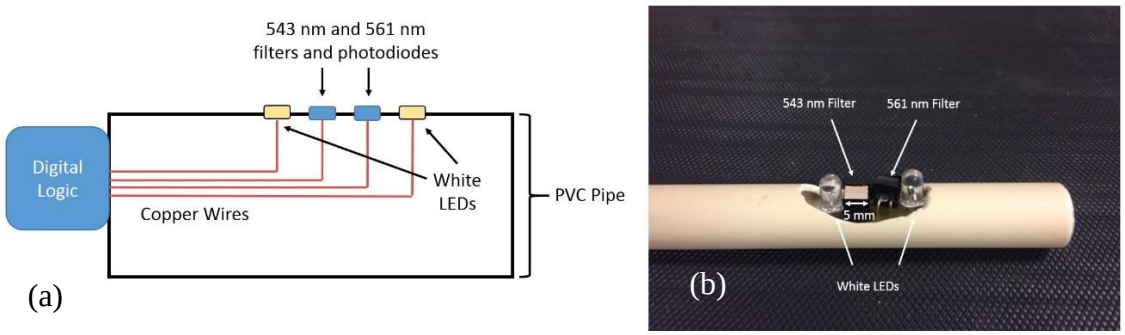

Figure 5: (a) Concept of the first intubation probe.
(b) First intubation probe prototype. The inner diameter of the probe is 15.8 mm; the outer diameter is 21.3 mm. The Semrock photofilter dimensions are 5x5x3 mm.

After building and testing the device and circuits to ensure functionality, the two photodiodes with their respective circuits were normalized. The normalization process ensures that any variance in emission, transmission, and detection at those wavelengths are accounted for, as each component (the emitter, fiber, and detector) has different responses at each wavelength. Prior to normalization, the dark voltages were measured to provide the reference. Normalization then involved taking measurements with the probe approximately 1 cm away from a standardized white reflectance target and adjusting the individual gain for each wavelength band such that the measured voltage minus the dark voltage were the same for each band.

After normalization, the test in the tracheal and esophageal tissues involved inserting the probe into the esophageal tube and tracheal tube and taking measurements of the output voltage. After subtracting the dark measurement from this reading, the values were then recorded and compared with Table 1 showing the expected results based on the tissue type.

Table 1: Predicted results of in vivo test.

	543 nm	561 nm
Esophagus	Higher Relative Voltage	Lower Relative Voltage
Trachea	Lower Relative Voltage	Higher Relative Voltage

2.2. Alternate Probe Design

Despite being able to fit the probe into the tissue as shown in Figure 7 in the results section, the fitment was extremely tight leading us to investigate a smaller probe design. Rather than having discrete components built on the probe itself, the smaller design utilized smaller LEDs to project white light onto the surface of the tissue and moved the filters and PDs outside the tube. The reflected returns were then collected by a plastic optical fiber that transmitted the signal to the filters and photodiodes corresponding to 543 nm, 561 nm, and 578 nm bands. The filters used were from Edmund Optics corresponding to 546 nm, 560 nm, and 577 nm. Each had full-width half-maximum band of 10 nm. The amplifier design was similar to the one shown in figure 4, but used OPA128L amplifiers instead of the LT1226. The OPA128L amplifier has a higher input resistance and lower noise overall than the LT1226, allowing for greater gain and less noise, required due to the reduced response from the signal collected and traveling through the fiber.

Testing and normalizing proceeded as above with the exception of using a ratio test of the voltages versus a straight comparison of the intensities. Using a ratio test reduced one of the normalizations steps. Instead of adjusting gain to

match relative voltages, the voltages were just measured and compared. A ratio of the voltages against the white reflectance targets were then compared with measurements of the tissue as shown in figure 8 below. This eliminates the need for any gain adjustment prior to taking measurements.

Ten measurements were taken of the spectrally flat surface to determine its average and standard deviation. The value of the 561/543 test was 3.07 with a standard deviation of .05; the value of the 561/578 was .578 and a standard deviation of .006. If the tracheal tissue ratios were greater than the spectrally flat ratios, then the tracheal tissue was properly detected. If the tissue ratios fell within the standard deviation or were less than the spectrally flat ratios, then the tissue was not detected.

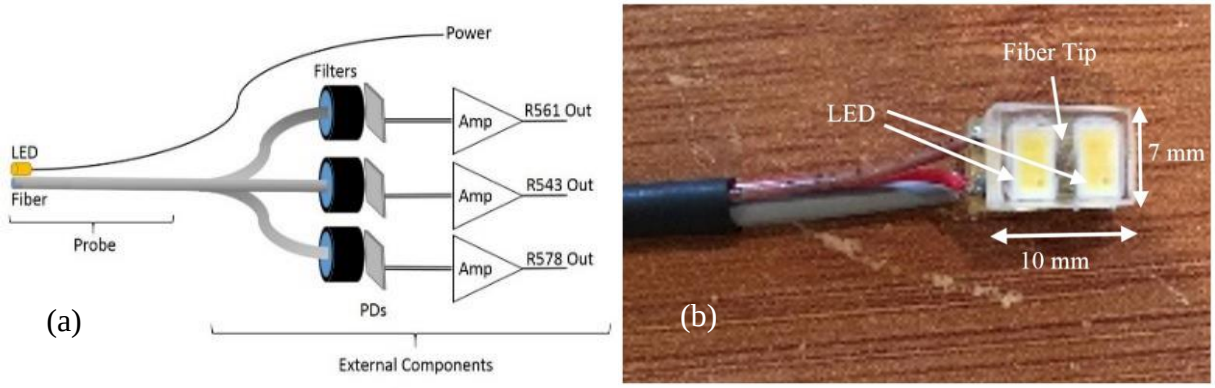

Figure 6: (a) Concept of the second probe design using filters from Edmund Optics.
(b) Probe showing the dimensions, LEDs, and optical fiber.

3. Results and Discussion

Initial detection results with the first probe design were inconclusive, likely related to tissue degradation due to a 24 hour delay in receipt of the tissue. However, we were able to successfully demonstrate that a probe made of discrete components, to include light source, filtering, and photodiodes could be fit into a tracheal tube as shown in Figure 7 below. Despite the tight and obvious impractical fitment, this design validates that with the use of surface mount components and special order filters (made to a much thinner thickness than the 5x5x3 mm ones used) a complete probe with both source and detection components can fit in a probe small enough to be used in a tracheal tube.

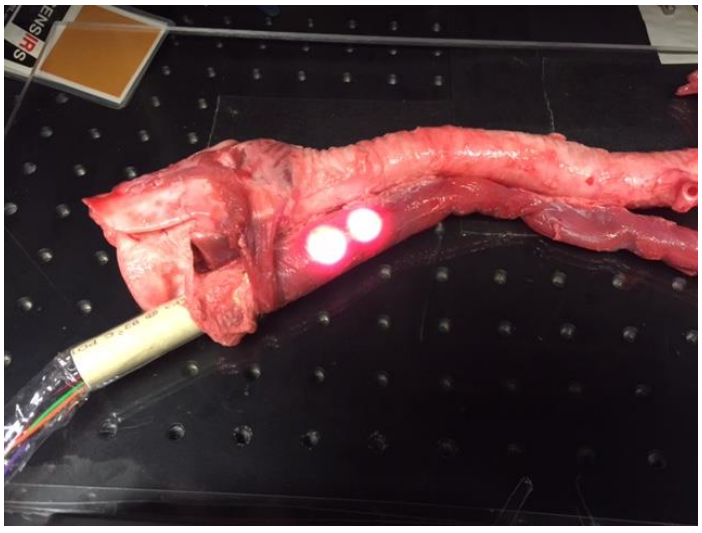

Figure 7: In vivo probe inserted into the esophagus of the pig.

The much smaller size of the second probe eased the fitment issue seen above. However, the probe exhibited a very weak response in the 543 nm band, with the signal just above the noise level. In examining the results from the ratio test below, where values at or less than the ratio measured against the white reflectance target are failed tests and values above are positive results, we see that the comparison of 543 nm to 561 nm had several failures. Alternatively, the comparison of the 578 nm to 561 nm, both of which signals were well above the noise, exhibited a 100% success rate.

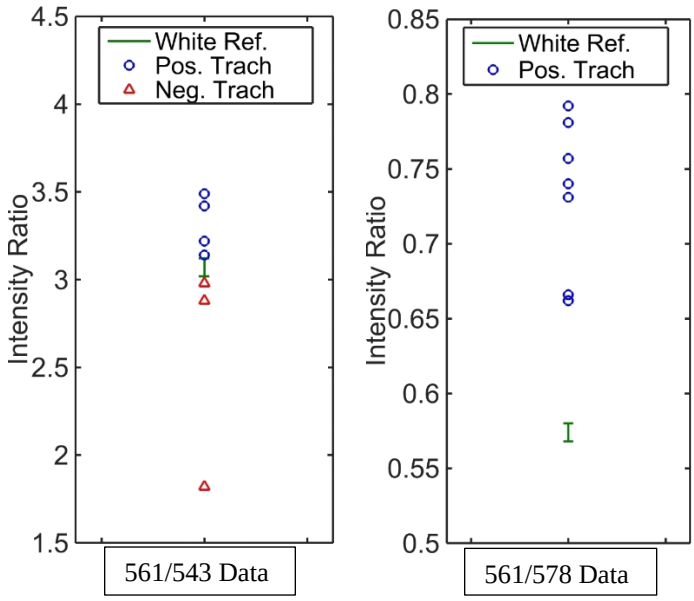

Figure 8: Ratio Test Results of 561/543 and 561/578

5. Conclusion and Future Work

Using hyperspectral imaging for in vivo detection needs further development. Some things to consider for future work is to create a smaller design that has a less substantial profile against the tissue to allow for optimal detection and reduce invasiveness of the device within the trachea and esophagus. Additionally, the presence of other bodily fluids like blood or mucus likely alters the reflectance spectrum of both tissues. Research of how to mitigate these other factors could influence future tests to improve detection.

Additional testing needs to be conducted using the discrete component device to identify the issue in its operability. It also requires further miniaturization. In using the fiber probe design, testing must be conducted on esophageal samples to determine if similar rates of success can be achieved. If the probe can positively detect the esophagus, then the design will be validated. Increasing the signal strength through the optical fiber and proper signal amplification can further refine the results to produce a more precise device.

6. Acknowledgments

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7. References

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