A Soft Sensor for Bleeding Detection in Colonoscopies

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Colonoscopies allow surgeons to detect common diseases, that is, colorectal cancer, ulcers, and other ailments. However, there is a risk of bleeding in the lower gastrointestinal (GI) tract while maneuvering endoscopes. This may be due to perforations, hemorrhaging, polyps, diverticula, or post-biopsy complications. Thus, it is essential for the surgeon to be able to detect bleeding at the site and evaluate the severity of blood leakage. Herein, a soft sensor that can detect the presence of blood at the bleeding site during colonoscopies is presented. The sensor consists of optical waveguides that interface with a microfluidic channel. Blood flow causes absorption and scattering of incident light that can be picked up by the optical sensing apparatus via light transmission through the waveguide. The surgeon can be alerted when bleeding occurs through a graphical user interface. The device is compact and measures only 1 mm thick. This allows the sensor to be circumferentially mounted onto a colonoscope at different locations. The sensor is able to record the presence of blood as an optical loss, rapidly detect the presence of blood under 100 ms as it enters the microchannel, and differentiate between gastric fluid and blood through changes in measured optical loss.

1. Introduction

In the United States alone, colorectal cancer was responsible for 53,200 deaths in 2020.[1] There are nearly 1.4 million new cases of colorectal cancer every year worldwide and 694,000 deaths.[2]

Colonoscopies are essential procedures that allow screening and timely detection of colorectal cancer and other serious ailments such as ulcerative colitis, diverticulitis, Crohn’s disease, and angiodysplasia.[3–6] It is of paramount importance that patients have access to colonoscopies procedures and thorough screening for such ailments. However, colonoscopic procedures pose certain safety risks to the patient due to the cumbersome navigation of the scope through the gastrointestinal (GI) tract. This may result in severe adverse events (SAE) such as perforation, bleeding, and splenic injury.[7] Bleeding occurs at a rate ranging between 6.5 and 23.1 cases per 10,000.[8] Specifically, navigation of the colonoscope in the sigmoid colon region poses a particularly high risk of bleeding, due to its tortuosity.[9] Bleeding in the lower intestine may be due to a variety of causes such as diverticula, perforation, hemorrhoids, colorectal polyps, and tissue biopsies.[6,10–13]

For example, perforation-related bleeding may be caused by the tip of the endoscope pressing against the wall or lateral force exerted by the looped colonoscope onto the side of the colon wall.[9] If bleeding occurs in the latter scenario, the surgeon may be unaware as only the distal tip of the endoscope is equipped with a camera. Thus, there is a pressing need for colonoscopic equipment to be able to detect any unwanted bleeding along the entire length of the device.

Current solutions to detect blood and other biological fluids are mainly geared toward microfluidic cytometry.[14] These lab-on-chip (LOC) detection systems focus primarily on analyzing the cell composition of biological fluids, that is, red blood cell counting or cell fluorescence detection,[15–17] instead of detection of whole biological fluids. Nevertheless, it is worth highlighting the few sensors that demonstrate the ability to detect whole blood, including sedimentation and blood clotting.[18–21]

A majority of these systems require expensive, high-precision optical equipment such as lasers, spectrometers, and optical fibers embedded in the device.[19,22] Cytometers also rely on pressurized tubing systems for hydrodynamic flow focusing in microchannels.[23–24] Thus, these sensors are limited by their structural rigidity and cumbersome optoelectronic setup. This makes it impractical to employ these sensors in clinical scenarios such as during colonoscopies, where there is continuous movement through tortuous regions and a requirement for real-time data collection (i.e., to detect bleeding). There has been progress toward designing LOC devices that incorporate soft optical sensing.[25–27] Numerous opto-fluidic sensors have successfully integrated polymeric waveguides to light across microfluidic
channels, in lieu of traditional optical fibers.\cite{28-31} Concurrently, there have been developments in the field of lab-on-fiber (LOF) devices, where specific materials can be embedded onto the distal tip of optical fibers to enable biological sensing.\cite{32} For instance, LOF sensors have demonstrated in-vivo imaging and fluorescence capabilities for endoscopies.\cite{33} However, LOF systems are limited to detection only at the distal tip of the fiber where light is emitted.\cite{34,35} This makes LOF devices unsuitable for GI bleeding detection when the bleeding occurs behind the distal tip of the colonoscope.

Overall, technologies for traditional interventional colonoscopies and robot-assisted endoscopies have mainly focused on endoscope shape sensing, tissue-force interaction, and colonic navigation to reduce occurrence of pain and SAEs.\cite{36-41} Examples of soft sensing technologies for colonoscopy include a soft robotic sleeve, with optical waveguides that can be mounted along a colonoscope to sense incident force between the instrument and the colon wall,\cite{42} distal tip force-interaction sensors,\cite{43} and endoscopic shape sensing using fiber Bragg grating (FBG) sensors.\cite{44} Ultimately, there appears to be a dearth of work seeking to integrate biological sensing capabilities onto colonoscopes.

This work presents a first proof of concept of a soft optical sensor to solve the issue of real-time blood detection in colonoscopies (Figure 1 and Movie S1, Supporting Information). The optical sensor is made of entirely soft polymers, which allows it to conform and wrap easily around the circumference of the colonoscope. The sensor does not rely on expensive optoelectronic equipment nor does it require pressurized tubing systems to sustain fluid flow in the microchannels. Instead, the sensor is able to efficiently intake fluid via capillary action. Optical detection relies on soft waveguides instead of a rigid fiber optic setup. The sensors can be integrated along the circumference of the colonoscope at multiple lengths, thereby allowing the surgeon to sense and locate bleeding during the procedure (Figure 1b,c).

2. Results and Discussion

This section outlines the soft sensor design, optical modeling, sensor fabrication, and control circuit. The section also describes the results of the blood sensor characterization tests comparing device sensitivity to different concentrations of blood, differentiation of other fluids, blood detection speed, and the in-vitro testing results.

2.1. Soft Sensor Design

The soft sensor is composed of two waveguides with a core cross section of 200 μm × 200 μm (Figure 2a and Movie S1, Supporting Information). Light is coupled into one waveguide via a 200 mm
core diameter optical fiber (BC03597-10-BL, OFS Fitel, LLC). The emitted light travels through the fluidic microchannel (for blood detection) via an in-plane polydimethylsiloxane (PDMS) micro-lens. The lens helps to focus the divergent light into the microchannel (Figure 2b). The light is then picked up by the opposing waveguide connected to another optical fiber to enable detection. The soft sensor is composed of a PDMS cladding (Dow Corning, Inc.) and a polyurethane core (Clear Flex 30, Smooth-On, Inc.). PDMS and Clear Flex 30 are both optically transparent elastomers with excellent light transmission properties.[45,46] The core of the waveguide (Clear Flex 30) has a refractive index (RI) \( n_3 = 1.486 \) and the cladding (PDMS) has an RI \( n_2 = 1.43 \). The difference in refractive indices allows the waveguide core to propagate light via total internal reflection. The waveguides allow total internal reflection at a maximum critical angle \( \theta_c \) of 74.2°, as calculated in Equation (1).

\[
\theta_c = \sin^{-1}(n_2/n_1)
\]  

(1)

The design of the sensor and the integrated lens was guided by optical simulations (see Section 2.2). Flow of blood via the micro-channel causes the light from the waveguides to be absorbed, leading to an attenuation in the monitored signal, that is, a drop in voltage from the measured baseline value. The waveguides are curved in a 90° bend, which allows the input and output optical fibers to be placed parallel to each other (Figure 2). This reduces the width of the sensor and allows it to be mounted onto a curved surface, that is, a colonoscope without significant flexion (Figure 1a). The waveguides have a radius of curvature (ROC) of 2 mm. A higher ROC than 2 mm was not considered as this would increase the width of the sensors and reduce the number of sensors that could be mounted onto the colonoscope. Similarly, waveguides with smaller ROC were not considered, as reducing the ROC incurs higher bend losses from radiation,[47] thereby having reduced transmission of light around the 90° bend.

The sensor measures 10 mm in width by 25 mm in length to allow mounting of up to three sensors (Figure 1a) radially around a colonoscope (colonoscope outer diameters range from 12 to 15 mm[48]). This in turn allows the sensors to detect bleeding at various locations around the circumference of the colonoscope. Furthermore, the sensor is designed to have a low profile (1 mm thick). This allows the sensor to be mounted onto the colonoscope without significantly increasing its outer diameter, which could hinder navigational capabilities and tool flexibility.

The sensor has a 5 mm inlet hole, which is connected to three microfluidic channels measuring 100 \( \mu \)m in width and 200 \( \mu \)m in height (Figure 2a). The channels widen to 1 mm and then bottleneck into a single channel passing by the waveguides. The narrow microchannels at the inlet prevent thick, viscous blobs of mucus or bile from entering the sensor. Blood is allowed to flow via capillary action. The channels then expand back into a wider outlet channel measuring 1 mm in width and 200 \( \mu \)m in height. The outlet is connected to 0.94 mm outer diameter (OD) flexible tubing (MicroRenathane 037, Braintree Scientific). The tubing is connected to a vacuum line which allows the blood to be suctioned out of the channel intermittently at the surgeon's command. Additionally, any accidental clogging of the channel can be resolved by flushing the channel using pressurized gas through the tubing.

2.2. Optical Modeling

A commercial optical ray-tracing software (Zemax OpticStudio) was used to guide and inform the design of the soft blood sensor. The primary motivation was to find a solution to optimally confine the divergent light rays emanating from the emitter waveguide into the microchannel. Divergent light can produce higher coupling losses at the interface between the waveguide and the microchannel (Figure 2). This may result in undesirable sensor behavior such as a lower signal-to-noise (SNR) ratio and reduced sensitivity.[49] Thus, an in-plane PDMS lens system was designed (Figure 3) to focus incident light from the input waveguide into the center of the microchannel. This allows the incident light to pass through the fluid and couple efficiently into the detector waveguide, thereby reducing any optical coupling losses that could affect sensor performance.

The selected lens configuration consisted of two PDMS micro-lenses separated by an air gap (Figure 3a). This allows collimation and focusing of the divergent light. The size of the lens was first approximated to fit in the space between the waveguide and the microchannel wall without excessively increasing the optical path length. A very long optical path length may lead to greater attenuation losses as the light travels through the PDMS–air media.

The first lens has a –200 \( \mu \)m ROC. The separation between the lenses is 170 \( \mu \)m and the second lens has a ROC of 100 \( \mu \)m. Alternatively, the air gap separating the two curved PDMS lenses can be considered to be a biconvex air lens in PDMS medium. Thus, the effective focal length of the lens system can be calculated using the thick lens equation for a lens in medium.

![Diagram showing layout and path of light passing through waveguide–microchannel interface and lens.](image-url)
\[
\frac{1}{f} = \left( \frac{n_l - n_m}{n_m} \right) \left[ \frac{1}{R_1} - \frac{1}{R_2} + \left( \frac{n_m - n_l}{n_l R_1 R_2} \right) d \right]
\]

(2)

where \( f \) is the effective focal length, \( R_1 \) is the ROC of the first surface, \( R_2 \) is the ROC of the second surface, \( n_m \) is the RI of the medium (PDMS \( \approx 1.43 \)), \( n_l \) is the RI of the lens (air \( \approx 1 \)), and \( d \) is the distance of the air gap. Thus, the theoretical effective focal length is calculated to be 267 \( \mu \text{m} \). This allows the divergent light to be collimated and then focused into the center of the microchannel to be collected by the opposite waveguide.

The entire sensor was then simulated as a nonsequential system. Light enters the first waveguide and propagates through the core via total internal reflection. Some of the light is lost as radiation along the curved bend. The resultant light exits the first waveguide and then is focused by the lens into the second waveguide as it passes through the microchannel, as illustrated in Figure 3c. The lens is able to confine a high percentage of the divergent light into the center of the channel to ensure the light enters the second waveguide across the channel. We also performed a nonsequential ray-tracing simulation of the sensor without the lens (Figure 3b). As seen in Figure 3d, the light exits the first waveguide and diverges as it enters the microchannels. This results in a greater percentage of light rays failing to enter the second waveguide, that is, an increase in coupling loss. The simulations were validated by fabricating physical sensors, one with the microlens and the other without the microlens. The sensor with the microlens had a higher average baseline voltage (2.1 V) compared to the sensor without the lens (0.91 V). The difference in baseline voltages demonstrates the ability of the lens in confining stray light to increase coupling efficiency. This provides the sensor with a larger sensing range and increases accuracy by mitigating background noise.[50]

2.3. Fabrication

The patterns for the microchannels and waveguides are fabricated via soft lithography (Figure 4 and Movie S1, Supporting Information). Two 3-in. silicon wafers are prepared: one patterned (height of features is 200 \( \mu \text{m} \)) with SU-8 2050 photoresist (Kayaku Advanced Materials, Inc.) and one blank. The height of the features is chosen to be 200 \( \mu \text{m} \) to ensure that the cross-sectional area of waveguide cores matches the diameter of the selected optical fiber and facilitates its insertion in the soft sensor. Current available SU-8 photoresists cannot reliably produce the required layer thickness in a single spin coat. Hence, the photoresist must be spin coated and soft baked in two successive layers to achieve the desired thickness. The first layer is 120 \( \mu \text{m} \) thick and the second layer is 80 \( \mu \text{m} \) thick. The microchannels and waveguides grooves are created using photolithography (Figure 4a) with Kapton masks (McMaster-Carr Supply Company). The Kapton masks have a thickness of 25.4 \( \mu \text{m} \) and are cut with the desired pattern using a 5 W solid-state laser (Matrix-355, Coherent, Inc.). Both wafers are placed in an evacuated chamber with an open vessel containing a few drops of trichloro(1H,1H,2H,2H-perfluorooctyl)silane (Sigma-Aldrich) for at least 3 h. PDMS is mixed in a 10:1 ratio and poured onto the patterned and blank wafers. The wafers are then spin coated at 150 rpm for 45 s, 5 s ramp to produce two 0.5 mm thick layers (Figure 4b). The PDMS layers are peeled off the wafer and subjected to plasma oxidation (Figure 4c) using a plasma etcher (M4L, PVA TePla America, Inc.). The recipe for the procedure is as follows: 60 W, \( \text{O}_2 \) gas, 200 sccm for 30 s. The layers are then pressed together to form a permanent bond and heated on a hot plate at 70 °C for 2 min (Figure 4d). The waveguide core is filled under vacuum and subsequent re-pressurization. Drops of the core material are placed above inlet holes on the empty waveguide channels and subjected to vacuum in a vacuum chamber at \(-100 \text{kPa} \) for 10 min. As the chamber is vented, the external air pressure pushes the core material into the evacuated waveguide channels. The sensor is then cured at 70 °C for 4 h in a convection oven. The sensors are sonicated in an ultrasonic deionized (DI) water bath (1510, Branson Ultrasonics Corporation, USA) to remove any dust particles clogging the microchannels. The sensor is then subjected to further plasma oxidation treatment in the plasma etcher at 70 W, \( \text{O}_2 \) gas, 200 sccm for 500 s to reduce hydrophobicity of the sealed PDMS microchannels.[51] This allows the oxygen plasma to enter the microchannels and smooth the inner surface to ensure that these surfaces stay hydrophilic for a longer period of time, thereby enabling capillary action of the blood from the inlet.

2.4. Control Circuit

The blood sensor is connected to an optoelectronic circuit via optical fibers (Figure 2a) that interface with the end facets of the soft optical waveguides. The optical fibers (BC03597-10-BL, OFS Fitel, LLC.) have a core diameter of 200 ± 4 \( \mu \text{m} \) and a buffer diameter of 230 \( \mu \text{m} \). The fibers are connected to a fiber-coupled light emitting diode (LED) and phototransistor. The LED (IF-E93, Industrial Fiber Optics) emits green light at 530 nm. The LED was chosen primarily due to the fact that the whole blood has absorbance peaks between 500 and 600 nm.[18,52] This allows the incident light to be largely absorbed by the blood in the channel. If another wavelength were used, that is, 630 nm, attenuation of light would be mainly due to scattering instead of absorption.[22]

The phototransistor (IF-D92, Industrial Fiber Optics) is set up as a simple voltage divider circuit with a 220 kΩ resistor in series. The output from the LED and phototransistor is connected to an Arduino Mega 2560 microcontroller. MATLAB (MathWorks, Inc.) is used to process and plot the acquired data. MATLAB is also used to create a graphical user interface that plots real-time loss data from the sensors connected to the system. In the event of blood leakage, the user is alerted of the specific sensor(s) that have encountered bleeding.

2.5. Soft Blood Sensor Characterization

This section covers the results of the testing of the soft sensor light intensity changes that result from varying concentrations of the blood analogue, different fluid inputs, and the detection speed of the sensors. A total of five sensors were tested for the light attenuation versus blood analogue concentration.
experiment. Similarly, another five sensors were fabricated and tested for the fluid differentiation tests.

2.5.1. Light Attenuation Versus Blood Analogue Concentration

The sensor was tested using varying concentrations of an artificial blood analogue (Carolina Biological Supply Company). This was done to ascertain the sensor’s ability to detect blood even if it is mixed with other bodily fluids. This is possible during colonic bleeding, where the blood may mix with GI fluid or mucus and thereby be diluted to a low concentration, which could potentially evade detection. The amount of light attenuated by the incident blood is dependent on the concentration. This can be represented by the Beer–Lambert law.

\[
A = \log_{10}\left(\frac{I_0}{I}\right) = \varepsilon lc
\]  

where \(A\) is the absorbance of the attenuating medium, \(I_0\) and \(I\) are the baseline power and output power, respectively, \(\varepsilon\) is the Molar absorption coefficient, \(l\) is the optical path length, and \(c\) is the Molar concentration. As the concentration of the blood increases, the absorbance of the solution increases. A larger amount of the incident light is absorbed while transmittance across the microchannel decreases. This can be measured as a drop in voltage across the photodiode detector at the end of the receiving waveguide. This is observed in Figure 5 wherein the optical loss increases as the volume concentration of the blood analogue increases.

The sensor is able to detect a range of concentrations from 20% to 100% and exhibits a fairly linear behavior. As the blood concentration increases beyond 60%, the experimental data show an increase in the standard deviation of the loss curve. This is because the corresponding output voltage measured by the photodiode is very low, thus a small variation in the output voltage between different devices that were tested can cause a large standard deviation in the subsequent loss. Nevertheless, this is relatively inconsequential as once the loss exceeds a threshold.

Figure 4. Soft sensor fabrication. a) SU-8 patterned onto wafer through photolithography. b) Polydimethylsiloxane (PDMS) spin coated onto patterned wafer. c) Blank and patterned PDMS is peeled off wafers and exposed to O2 plasma at 60 W, 30 s. d) PDMS layers are bonded together post plasma modification. e) Waveguide channels are filled with core material (Clear Flex 30) under vacuum. f) Sensor is cured at 65 °C for 4 h. g) Sensor is subjected to second plasma modification at 70 W, 500 s.

Figure 5. Soft bleeding sensor characterization of Blood Concentration (v/v) versus Optical Loss (dB). The solid line is the mean value and the shaded area represents the standard deviation computed on five prototypes.
value due to bleeding (i.e., 2 dB), the sensor will detect the presence of blood and the surgeon can be alerted via the graphical user interface (see Movie S1, Supporting Information). Thus, the surgeon is informed of potential bleeding regardless of the final loss value as long as it exceeds the defined threshold.

2.5.2. Differentiation between Blood and Other Fluids

Colonoscopy preparation involves evacuating the lower GI tract of any food particulates and fecal matter. However, there is still the presence of GI fluids in the colon. Thus, it is required that the blood sensor be able to differentiate between blood and gastric fluid. A gastric fluid analogue was prepared using 0.2% (w/v) NaCl solution in water. The solution was colored with yellow food dye to replicate gastric fluid that is combined with bile and mucus, which commonly occurs in the GI tract. A total of five sensors were manufactured for this test, and each sensor was tested ten times. Figure 6 shows the average loss before and after either the gastric fluid or blood enters the microchannel.

As seen in Figure 6, the presence of the gastric fluid in the sensor channel causes a gain in voltage. This is calculated as a negative loss as the output voltage is higher than the initial baseline where air is the medium present in the microchannel. The resulting attenuation appears to be minimal due to absorption and scattering. Additionally, as the gastric fluid analogue enters the waveguide–microchannel interface, intensity of light across the microchannel increases. This is because the reflection losses are lower at the PDMS–gastric fluid interface compared to the PDMS–air interface. Similar behavior has been noted in other PDMS–material interfaces where the RI difference between the two media is large. The reduction in reflection losses is seen as an increase in light transmission across the channel. In contrast, the blood analogue (100% concentration) causes a voltage drop (Figure 6), which is interpreted as an increase in loss. It is worth noting in Figure 6 that the initial loss for the line graph representing blood is not zero when the microchannel is empty. This is due to the fact that the blood analogue tints the microchannel as it passes through. This causes the new baseline voltage to be lower than the original baseline in subsequent trials. When compared to the baseline before any of the blood analogue tinted the channel, this is seen as a marginal loss (≈0.8 dB). This could be a potential limitation when the sensor detects blood at multiple instances and the channel is suctioned clear. After the first instance of detection, the tinting of the channel with blood will cause the sensor to show a slight loss. However, this can be adjusted for in the control program, where the baseline is recalculated after blood is detected and cleared from the channel.

2.5.3. Detection Speed

The sensors ability to detect blood via capillary action was evaluated by dropping the blood analogue into the sensor inlet after exposure to air at varying intervals of time ranging from 0 min to 4 h. This was done to ascertain if prolonged exposure to air resulted in a loss of hydrophilicity within the microchannels, which could potentially decrease the speed of capillary action and increase detection time. The detection time was defined as the time it takes for the blood analogue to travel from the inlet to the waveguide–microchannel interface. The distance between these features measures 2.86 mm (Figure 2a). The detection time was calculated by video capturing the flow of blood from the inlet to the waveguide–microchannel interface using a camera (D7500, Nikon) connected to a microscope (Stemi 508, Zeiss Microscopy). The camera was set to 60 frames per second. The number of frames were analyzed from the instant that the blood enters the inlet and reaches the waveguide interface. The average detection time for the 4-h test period is 87 ±53 ms. The detection speed is rapid which ensures that the sensor detects blood almost immediately after contact with a bleeding spot even after exposure to air for 4 h. Most colonoscopies last anywhere between 15 and 60 min, with an average non-polyp procedure time of 36 min. Thus, the hydrophilicity of the sensor when exposed to air is retained well beyond the required time period for most colonoscopies. This ensures that the sensor performance and detection speed do not degrade over the course of the colonoscopy, which could affect bleeding detection. Possible explanations for the variation in detection speed during the test may be air bubbles being trapped in the microchannels or minor obstructions in the microchannel such as particulate matter.

2.6. In-Vitro Testing

The sensor was subjected to in-vitro testing in a mock colon to evaluate its performance in a simulated surgical setting (Figure 7 and Movie S1, Supporting Information). A mock colon (50 mm in diameter) was manufactured using films of thermoplastic elastomer (TPE) (Stretchlon 200, FibreGlast, USA) that were heat bonded together using an impulse sealer. The hollow TPE cylinders serve as the walls of the colon and were glued on to circular acrylic holders for structural support. A perforated TPE pouch was heat bonded internally to the top of the TPE colon wall.
(see schematic in Figure 7). The pouch can hold a small volume of blood analogue, and upon contact, blood drips through the perforations. The perforated pouch simulates bleeding scenarios such as perforation due to excessive force, or contact-induced bleeding of diverticula due to the colonoscope. Three blood sensors were fabricated and bonded onto a fabric sheet using silicone adhesive (Sil-Poxy, Smooth-On), as shown in Figure 7c. Once the sensors are securely bonded, the fabric is hand sewn into a sleeve and then mounted onto a mock colonoscope tube, 12 mm outer diameter (McMaster-Carr Supply Company). Optical fibers were inserted into the ends of the blood sensor waveguides and connected to the optoelectronic circuit. As the colonoscope navigates the TPE colon, it contacts the perforated pouch and causes leakage of blood the TPE colon wall. The blood passes into the sensor microchannel via capillary action and the sensor detects the presence of blood as it attenuates the incident light from the waveguides at the microchannel interface. Results from the in-vitro tests are shown in Figure 8. One of the three sensors detects blood (around the 20 s mark) and the loss curve rapidly increases.

3. Conclusion

This work presents a first proof of concept of a soft sensor that can detect bleeding events during colonoscopy. The sensor can be easily and rapidly manufactured using low-cost soft polymers and integrated onto a colonoscope. The device is disposable and can be replaced between colonoscopy procedures. Additionally, the simple optoelectronic circuitry allows multiple sensors to be integrated at a low overall cost. This is advantageous compared to traditional LOC systems that utilize expensive laser diodes and complex data acquisition systems. The small size and low vertical profile (1 mm total thickness) allows the sensor to be attached to the endoscope without significantly increasing its outer diameter or hampering navigation capabilities and instrument flexibility. The compact width of the sensors allows for three sensors to be mounted along the radial section of the colonoscope. This allows bleeding detection around the entire circumference of the colonoscope. Further, this allows identification of blood leakage even if the bleeding location is far away from the distal tip (camera) of the colonoscope.

The operability and functionality of the sensors have been validated in this work, showing the sensor’s ability to differentiate between different intestinal fluids, identify the concentration of the blood leakage, and perform in a simulated surgical setting. The sensor is able to detect varying concentrations of the blood analogue ranging from 20% to 100% and exhibits a corresponding average loss ranging from 1 to 7 dB. The sensor also exhibits a voltage gain in the presence of gastric fluid and a voltage loss in the presence of blood, thereby enabling differentiation. The device incorporates an in-plane microlens to focus divergent light exiting the waveguides into the microchannel. This was validated by using an optical tracing modeling software and comparing the average baseline values for the sensor prototypes with the lens (2.1 V) and without the lens (0.91 V). The sensor shows sustained capillary action and hydrophilic behavior well beyond a few hours with a detection time of 87 ± 53 ms. This could allow the sensor to be stored in a vacuum-sealed bag or in water to prevent exposure to air, thereby allowing the sensor to be opened and used in the operative room at a later period of time. Furthermore, the vacuum suctioning tubing will allow the bodily fluids to be collected for further analysis, that is, cancer screening.

Future work aims to further reduce the size of the sensor, including reducing the diameter of the waveguides and thickness.
of the cladding. This can enable a high density of sensors to be attached to the colonoscope. Numerous triplets of sensors can be potentially mounted along varying lengths of the colonoscope to enable blood detection along the whole length of the device. Ultimately, we aim to further validate the functionality of the sensor by testing the apparatus in-vivo, that is, on a live animal porcine model.

Supporting Information

Supporting Information is available from the Wiley Online Library or from the author.

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Conflict of Interest

The authors declare no conflict of interest.

Data Availability Statement

The data that support the findings of this study are available from the corresponding author upon reasonable request.

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[1] R. L. Siegel, K. D. Miller, A. Jemal, Cancer J. Clin. 2020, 70, 7.
[2] H. J. Bonjer, C. L. Deijen, G. A. Abis, M. A. Cuesta, M. H. van der Pas, E. S. de Lange-de Klerk, A. M. Lacy, W. A. Bemelman, J. Andersson, E. Angenete, J. Rosenberg, A. Fuerst, E. Haglind, N. Engl. J. Med. 2015, 372, 1324.
[3] R. C. Langan, P. B. Gotsch, M. A. Krafczyk, D. D. Skillinger, Am. Family Phys. 2007, 76, 1323.
[4] S. Baurn, C. A. Athanasoulis, A. C. Waltman, J. Baldabini, R. H. Schapiro, A. L. Warshaw, L. W. Ottinger, Am. J. Roentgenol. 1977, 129, 789.
[5] P. V. Rasmusson, F. Dalgaard, G. H. Gislasen, A. Brandes, S. P. Johnsen, E. L. Grove, C. Torp-Pedersen, L. Dybro, L. Harboe, A.-M. B. Münster, L. Pedersen, P. Blanche, J. L. Pallisgaard, M. L. Hansen, Eur. Heart J. 2022, 43, e38.
[6] R. Cirocchi, V. Grassi, D. Cavaleri, C. Renzi, R. Tabola, G. Poli, S. Avenia, E. Farinella, A. Arezzo, N. Vettoretto, V. D’Andrea, G. A. Binda, A. Fingerhut, Medicine (United States) 2015, 94, 1710.
[7] S. T. Kothari, R. J. Huang, A. Shaukat, D. Agrawal, J. L. Buxbaum, S. M. Abbas Fehami, D. S. Fishman, S. R. Gurudu, M. A. Khashab, L. H. Jamil, T. L. jue, J. K. Law, J. K. Lee, M. Naveed, B. J. Qurnsey, M. S. Sawhney, N. Thosani, J. Yang, J. M. DeWitt, S. Wani, Gastrointestinal Endoscopy 2019, 90, 863.
[8] M. Laanani, J. Coste, P. O. Blotiére, F. Carbonnel, A. Weill, Clin. Gastroenterol. Hepatol. 2019, 17, 719.
[9] Y. Jung, Clin. Endosc. 2019, 53, 29.
[10] S. M. Kavic, M. D. Basson, in Surgical Treatment: Evidence-Based and Problem-Oriented (Eds.: R. G. Holzheimer, J. A. Mannick), Zuckschwerdt, Munich 2001.
[11] A. U. R. Jehangiri, R. Gul, R. Hadayat, A. N. Khan, Zabiullah, L. Khursheed, J. Ayub Med. College, Abbottabad: JAMC 2017, 29, 468.
[12] H. S. Kim, T. I. Kim, W. H. Kim, Y.-H. Kim, H. J. Kim, S.-K. Yang, S.-J. Myung, J.-S. Byeon, M. S. Lee, I. K. Chung, S.-A. Jung, Y. T. Jeen, J. H. Choi, K. Y. Choi, C. H. Han, S. Dong, J. S. Song, Am. J. Gastroenterol. 2006, 101, 1333.
[13] D. C. Rockey, Gastroenterology 2006, 130, 165.
[14] J. Godin, C.-H. Chen, S. H. Cho, W. Qiao, F. Tsai, Y.-H. Lo, J Biophotonics 2008, 1, 355.
[15] Y. Zhao, Q. Li, X. Hu, Y. Lo, Biomicrofluidics 2016, 10, 6.
[16] Y. Zhao, Q. Li, X. M. Hu, D. F. Yang, Photonics Lasers Med. 2013, 2, 51.
[17] D. Huh, W. Gu, Y. Kamotani, J. B. Grotberg, S. Takayama, Physiol. Meas. 2005, 26, 3.
[18] H. Ashiba, M. Fujimaki, K. Awasu, T. Tanaka, M. Makishima, Sens. Bio-Sens. Res. 2016, 7 121.
[19] N. Elshikhi, H. Bakhtiari, J. Phys.: Conf. Ser. 2018, 1027, 1.
[20] N. Taparia, K. C. Platten, K. L. Anderson, N. J. Sniadecki, AIIP Adv. 2017, 7, 10.
[21] C. Cab-Cauch, J. J. Alvarado-Gil, O. L. Leãnos-Castãneda, Rev. Sci. Instrum. 2006, 77, 4.
[22] M. Ueda, K. Ishikawa, C. Jie, M. Sanae, Y. Tounra, Opt. Lasers Eng. 1994, 21, 307.
[23] M. Rosenauer, W. Buchegger, I. Finoulst, P. Verhaert, M. Vellekoop, Microfluidics Nanofluidics 2011, 10, 761.
[24] S. H. Huang, W. H. Tan, F. G. Tseng, S. Takeuchi, J. Micromech. Microeng. 2006, 16, 2336.
[25] J. Wang, J. Dong, Sensors (Switzerland) 2020, 20, 1.
[26] D. A. Chang-Yen, R. K. Eich, B. K. Gale, J. Lightw. Technol. 2005, 23, 2088.
[27] F. Song, D. Pagliero, C. A. Meriles, S.-W. Seo, Opt. Eng. 2013, 52, 044404.
[28] H. Hosseinikhannazer, L. W. Kostiuk, J. N. McMullin, in Photonics North 2008, Society of Photo-Optical Instrumentation Engineers (SPIE) Vol. 7099, 2008, p. 70990H.
[29] C. L. Bliss, J. N. McMullin, C. J. Backhouse, Lab Chip 2007, 7, 1280.
[30] P. Fei, Z. Chen, Y. Men, A. Li, Y. Shen, Y. Huang, Lab Chip 2012, 12, 3700.
[31] M. Fleger, D. Siepe, A. Neyer, IEEE Proc. Nanobiotechnol. 2004, 151, 159.
[32] M. Pisco, A. Cusano, Sensors (Switzerland) 2020, 20, 1.
[33] G. Oh, E. Chung, S. H. Yun, Opt. Fiber Technol. 2013, 19, 760.
[34] F. Tian, S. Sukhishvili, H. Du, Lab-on-Fiber Technol. 2015, 56, 315.
[35] A. Ricciardi, M. Consales, G. Quero, A. Crescitelii, E. Esposito, A. Cusano, Opt. Fiber Technol. 2013, 19, 772.
[36] M. Chauhan, J. H. Chandler, A. Jha, V. Subramanian, K. L. Obstein, P. Valdastr, Front. Robotic. AI 2021, 8, 1.
[37] J. W. Martin, B. Scaglioni, J. C. Norton, V. Subramanian, A. Arezzo, K. L. Obstein, P. Valdastr, Nat. Mach. Intell. 2020, 2, 595.
[38] G. Ciuti, K. Skonieczna-Zydecka, W. Marlicz, V. Iacovacci, H. Liu, D. Stoyanov, A. Arezzo, M. Chiurazzi, E. Toth, H. Thorlacius, P. Dario, A. Koulaouzidis, J. Clin. Med. 2020, 9, 6.
[39] C. Shi, X. Luo, P. Qi, T. Li, S. Song, Z. Najdovski, T. Fukuda, H. Ren, IEEE Trans. Biomed. Eng. 2017, 64, 1665.
[40] S. Russo, T. Ranzani, H. Liu, S. Nefti-Meziani, K. Althoefer, A. Menciassi, Soft Robot. 2015, 2, 146.
[41] S. Sareh, Y. Noh, M. Li, T. Ranzani, H. Liu, K. Althoefer, Smart Mater. Struct. 2015, 24, 1.
[42] M. McCandless, A. Gerald, A. Carroll, H. Aihara, S. Russo, IEEE Robot. Autom. Lett. 2021, 6, 5292.
[43] T. Watanabe, T. Iwai, Y. Fujihira, L. Wakako, H. Kagawa, T. Yoneyama, Sensors (Switzerland) 2014, 14, 5207.
[44] Y. Lu, B. Lu, B. Li, H. Guo, Y. H. Liu, IEEE Robot. Autom. Lett. 2021, 6, 4835.
[45] D. K. Cai, A. Neyer, R. Kuckuk, H. M. Heise, Opt. Mater. 2008, 30, 1157.
[46] H. Bai, S. Li, J. Barreiros, Y. Tu, C. R. Pollock, R. F. Shepherd, Science 2020, 370, 848.
[47] I. Papakonstantinou, K. Wang, D. R. Selviah, F. A. Fernández, Opt. Express 2007, 15, 669.
[48] L. Manfredi, Front. Robot. AI 2021, 8, https://doi.org/10.3389/frobt.2021.705454.
[49] J. Tang, G. Qiu, X. Cao, Y. Yue, X. Zhang, J. Schmitt, J. Wang, Lab Chip 2020, 20, 2334.
[50] S. Camou, H. Fujita, T. Fujii, Lab Chip 2003, 3, 40.
[51] S. H. Tan, N. T. Nguyen, Y. C. Chua, T. G. Kang, Biomicrofluidics 2010, 4, 3.
[52] S. Liu, Z. Deng, J. Li, J. Wang, N. Huang, J. Biomed. Opt. 2019, 24, 1.
[53] N. Chronis, L. P. Lee, Lab Chip 2004, 4, 125.
[54] G. S. Raju, V. Vadyala, R. Slack, S. G. Krishna, W. A. Ross, P. M. Lynch, R. S. Bresalier, E. Hawk, J. R. Stroehlein, Cancer Med. 2013, 2, 391.