I. Executive Summary

In the United States, 2.2 million adults suffer from severe allergen-induced asthma (Appendix E), which causes airflow obstruction in the lungs due to inflammation[81]. This chronic respiratory disorder is currently detected with a peak flow meter, which relies entirely upon patients to monitor their breathing[69]. However, patients often underestimate the severity of their asthma and neglect to use their peak flow meter, resulting in hospitalizations due to a disorder that is preventable if detected early enough[26]. In order to improve patient quality of life and lighten the economic burden of asthma, we aim to engineer a safely implantable device requiring minimal patient involvement that detects and notifies adults who are suffering from a severe, allergen-induced asthma attack.

Our team has designed an automated device, consisting of an implantable stent and external interface, that continuously monitors bronchoconstriction and notifies patients when the radial pressure of the lobar bronchus is indicative of an allergen-induced asthma attack. A modified stent supporting a micro-pressure sensor and electronics is inserted bronchoscopically into an adult's lobar bronchus. If the pressure sensor detects a 20% decrease in bronchial diameter, the external monitor adhered to the patient's chest alerts the patient with ample time to administer medication, thereby preventing further escalation or hospitalization. We estimate that our device detects asthma within 20 minutes of allergen exposure and costs approximately $1,300. The device requires charging approximately every 6 years, but can feasibly function through a patient's lifetime if properly maintained.

With a $16,000 net benefit to the patient over a 6-year period, the Asthma ConStent is the best possible option for asthma attack detection in severely asthmatic patients. Considering the minimal invasiveness of implantation, high durability, and quick notification time, our product allows asthma patients to lead fuller, more productive lives, free from the worry of experiencing an uncontrollable attack.

II. Introduction

In the United States, 27,000 adults miss work due to asthma every day, resulting in nearly 15 million lost or less productive workdays each year[9]. The Center for Disease Control defines asthma as a chronic disorder with episodes or attacks characterized by periods of reversible airflow obstruction[81]. These attacks may vary from mild to severe and life threatening.

The estimated annual cost of asthma in the United States, including medical expenses, loss of productivity from missed school or work days, and premature deaths, totals to $56 billion[81].

Currently, a peak flow meter is the most commonly implemented device for in-home long-term asthma observation. The peak flow meter is a device that can measure Peak Expiratory Flow (PEF) rate, which is a person's maximum rate of forced expiration[69]. The peak flow meter requires the patient to track their PEF and compare output values to determine if he or she is experiencing an attack. This method relies upon the patient to monitor his or her asthma. Studies show that adults with moderate to severe asthma are very poor at taking daily measurements or gauging the severity of an asthma attack[25]. Ultimately, the peak flow meter is an insufficient method of monitoring asthma due to the unreliability of patients.

Our team responds to the lack of an adequate asthma monitor with a safe, implantable device that automatically detects and notifies a patient of the onset of an attack before symptoms worsen and require hospitalization. Our team's device, named the Asthma ConStent, is an implantable modified stent with a capacitive pressure sensor that detects bronchoconstriction. An external component alerts the patient of an attack. With this device, we hope to make asthma a more manageable disorder as well as reduce costs.

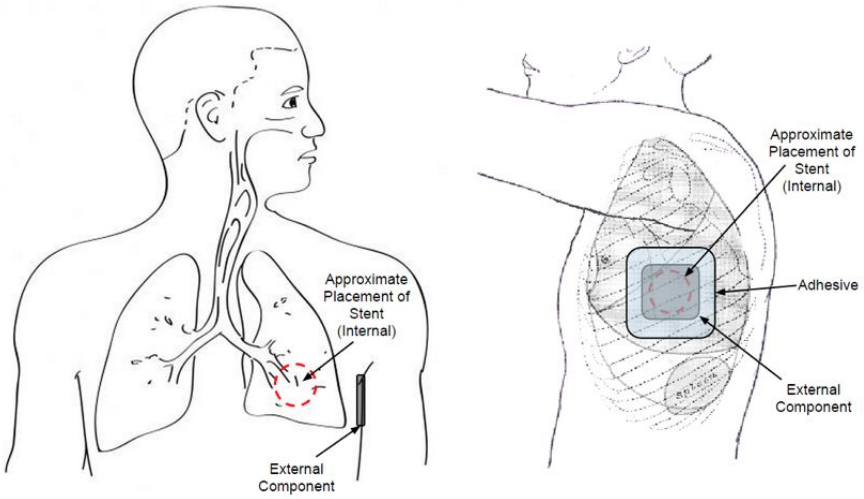

Figure 1: Placement of Asthma ConStent

Adapted from http://www.sewlavie.com/human-lungs-diagram/human-lungs-diagram-blank/#main

III. Target Population

We have chosen to define an asthma attack to be the reversible constriction of the bronchi. We define our population to be male and female adults, 18 years or older, who have severe, allergen-induced asthma. We have chosen to address allergen-induced asthma because exercise-induced asthma is inherently more preventable. Severe asthma is defined by the need for

daily intake of corticosteroid medications. Additionally, our population is defined to be above the poverty line, with access to adequate insurance and specialized health care facilities.

IV. Engineering Goals

The viability and efficacy of our proposed method was evaluated based on several design criteria and target goals. After initial screening, our proposed devices were all implantable and required medical procedures, so our design criteria are oriented towards implantable devices. In order of decreasing importance, our design criteria are: Level of Patient Involvement, Notification Time, Complexity of Implantation, Durability, and Cost. An ideal device would meet each target goal for each criteria, but a more realistic device simply fulfills the majority of the goals.

Design Criteria

Level of Patient Involvement: The level of patient involvement is defined to be how much time the patient must spend maintaining and actively interacting with the device. This includes factors such as how often the battery must be recharged and how much time and effort the device requires from the patient. This is important in creating our device because patients often fail to complete such tedious protocols. An ideal device is fully automated and requires no patient involvement or maintenance.

Notification Time: Notification time is defined as how long after allergen exposure that the symptom the device measures appears. Our team would like to minimize notification time to maximize the patient's time before an asthma attack becomes severe; an ideal device has a notification time of 20 minutes or less after allergen exposure.

Complexity of Implantation: This is defined as the discomfort and inconvenience the patient would experience as a result of the implantation procedure. Levels of complexity were defined in terms of procedure time, degree of anesthesia required, requirement of stitches, and length of subsequent hospital stay. An ideal implantation procedure is an outpatient procedure that takes 30 minutes or less, with only local anesthesia necessary, and no stitches.

Durability: The durability of the device accounts for lifetime and is quantified by frequency of implant replacement. Since any type of implantation procedure will have potential risks, it is important that our device does not require frequent replacement. An ideal device would not require any replacements or repairs after implantation.

Cost: The cost is defined as the cost of the device itself; our team did not consider cost of the procedure in this category because all proposed methods involve baseline surgical costs. Our team

assumed the surgical price of all proposed devices are comparable on the magnitude of thousands of dollars. An ideal device, with all components summed, would cost below $500.

These criteria were taken into account in a scoring matrix to evaluate the best possible device.

B. Scoring Matrix

		Bronchial Stent with Infrared Diameter Sensor		Bronchial Stent with Pressure Sensor		Flow Bronchial Stent		Catheter		Blood Stent	
Evaluation Criteria	**Weight**	rating	weight score	rating	weight score	rating	weight score	rating	weight score	rating	weight score
Level of Patient Involvement (Charging batteries, refilling chemicals, etc.)	**0.25**	3	0.75	4	1	3	0.75	1	0.25	1	0.25
Notification Time	**0.25**	5	1.25	5	1.25	3	0.75	2	0.5	2	0.5
Complexity of Implantation (invasiveness, asleep)	**0.2**	4	0.8	5	1	4	0.8	3	0.6	3	0.6
Durability/Frequency of Implant Replacement	**0.15**	4	0.6	4	0.6	4	0.6	2	0.3	4	0.6
Cost (of device, not including procedure)	**0.15**	3	0.45	4	0.6	4	0.6	3	0.45	4	0.6
Total Score	**1**		**3.85**		**4.45**		**3.5**		**2.1**		**2.55**

The Bronchial Stent with Pressure Sensor, renamed the Asthma ConStent, was chosen as the best device as it meets several of the target goals, including the notification time and complexity of implantation. The device approaches our ideal goal for level of patient involvement and durability. Although it does not perfectly meet all target goals, it is the most appealing option to patients for automated asthma monitoring.

The ConStent can be divided into 4 sections: a modified stent, a pressure sensor, electronics, and external notification interface. The stent acts as the framework for holding the pressure sensor, which is modeled after a sensor found in a cutting-edge intraocular implant[65]. After signal changes with an application specific integrated circuit, the information on constriction is sent to a small external component containing the circuitry for notification and powering the entire device, which is adhered to the skin.

V. Bronchoconstriction within the Asthma Timeline

A fundamental basis of the ConStent is that bronchoconstriction is a primary symptom and

effective indicator of an asthma attack[105]. We determined the efficacy of this indicator through studies examining asthma diagnosis based on different symptoms; these studies presented 84.6% sensitivity and 80.5% specificity as a predictor of asthma. The only better indicator of asthma is wheezing, which is a secondary symptom that is caused by bronchoconstriction[91,5,] Therefore, because our device relies upon a symptom that is present in every patient and for every attack, it measures the best possible indicator of an asthma attack.

For complete bronchial diameter calculations and justifications for the numbers presented here, see Appendix F. Before an asthma attack occurs, the average diameter of the generation 2 bronchus is 10.1 mm[64]. After allergen exposure, the patient will begin to experience the Early Phase Response, or EPR, which lasts approximately 2-5 hours [22,92]. During this period, bronchoconstriction occurs and maximizes at 30% for cases of moderate asthma[22], but can reach 69% in cases of fatal asthma[44]. Patients may administer short-term corticosteroids, that begin reducing bronchoconstriction within a few minutes and are effective for 4-6 hours[70].

A late phase response, or LPR, presents in about 60% of asthmatics and involves the immune system response of releasing inflammatory cells, causing bronchoconstriction as well[22]. The LPR begins 3-5 hours after exposure and reaches a maximum 6-12 hours after exposure; the LPR may persist for up to 24 hours[22,88]. The patient may medicate to dilate the bronchioles.

Our device notifies the patient when constriction reaches a threshold of 20%, which correlates to an average bronchial diameter of 8.08 mm. This threshold is high enough to avoid misdiagnosis due to bronchoconstriction from coughing, which can at most cause approximately 16% constriction in large airways[61]. Based on mouse models, we estimate that our device detects the attack within 20 minutes after exposure to the allergen. Considering that only 13% of severe asthma attacks develop over the course of an hour while the majority of them develop within 24 hours[112], the patient is notified at least 40 minutes before the asthma attack becomes severe. Based on EPR length, the patient is potentially able to cut an asthma attack short by at least 1-4.5 hours, during which bronchoconstriction could have exacerbated even further.

VI. Component Details

A. Stent

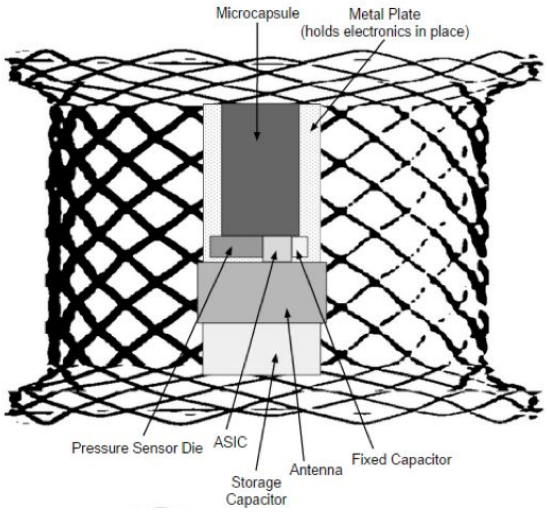

Figure 2: Modified Stent with Electronics

The principal component of our device is a self-expanding, metallic stent (SEMS) located in the upper lobar bronchus, designated as generation 2 in Weibel's model(#ref). This acts as the frame for our device, serving as a vessel for the pressure sensor and transmission technology. A diagram of the stent can be seen in Figure 2 above. Currently, bronchial stents can be inserted in the upper lobar bronchus to open narrowed airways[103]. The technology of a stent provides several advantages; it is widely implemented, proven to endure, and is easily manufactured in a variety of sizes.

Our stent has a lattice or mesh design cut from a Nitinol, or nickel-titanium alloy, tube, modeled after the Protege™ EverFlex esophogeal stent. The choice in metal material allows for a more permanent implantation while preserving a balance between rigidity and durability. Based on the data collected from radial compression tests and flexural rigidity tests, the Protege was selected because it had the highest stent stiffness in the radial direction[108]. Although the majority of the stent is made of nitinol mesh, one rectangular section consists of a metal plate housing the electronics which measure bronchial constriction.

We have modified our stent to make it safer and mitigate complications. The stent is fully covered in a biocompatible polyurethane coating in order to decrease tissue ingrowth or granulomas, around the stent[3] . Outer polyurethane coating is commonly used for its valuable properties such as elasticity, durability, and acceptance into the body during healing[68]. The coating also facilitates the removal and replacement of our device in case of complication or breakdown. To prevent the complication of stent migration caused by bronchial movement, we have designed an anti-migration technology of flared ends based on tests that determine the most effective anti-migration systems. These mechanisms are proven to deter migration for fully covered self-expandable nitinol stents used

on benign biliary structures[34], and we foresee similar circumstances and efficacy within the bronchus. The flared ends add 1 mm of length to each end of the stent and the diameter is 2 mm wider than the middle diameter of the stent[13]. Our stent is located in generation 2, as opposed to the first generation or primary bronchus, to hinder possible ejection of the stent[103].

We define a total stent length of 9 mm as long enough to hold the pressure sensor technology while allowing for sufficient contact with inner bronchi wall. Our device is significantly shorter than the lobar bronchus and thus avoids possible complications arising from curvature in the region. The average stent diameter measures 11.1 mm in the middle. Self-expanding stents are commonly held in place by the radial force they exert on the inner bronchial wall. Thus, our stent contracts to the diameter of the bronchi: 10.1 mm upon implantation[64]. For a complete list of calculations regarding diameters, see Appendix F.

B. Pressure Sensor

The pressure sensor is the main component of the Asthma ConStent for detecting bronchoconstriction. The Capacitive Pressure Sensor (CPS) is embedded within the center of the this metal plate, as seen in Figure 2 and Appendix D. We will be using the Protron Mikrotechnik Capacitive Pressure Sensor[83]. The CPS dimensions are 1.2 mm x 0.6 mm x 0.5 mm, and it is situated so that the sensing mechanism is facing the bronchial wall. This orientation allows the CPS to detect bronchial constriction by measuring the pressure the bronchus exerts on the stent. One CPS is sufficient for the device, as radial pressure can be detected from any side of the stent.

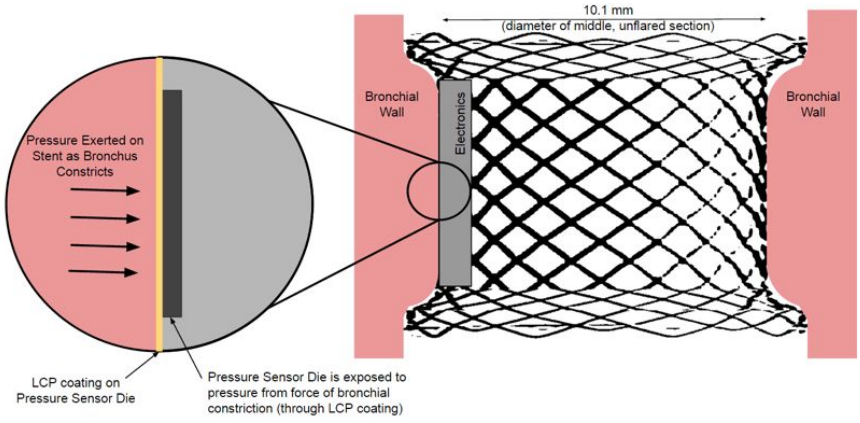

Figure 3: Orientation of CPS

The CPS uses capacitors to measure absolute pressure. The basic mechanism of the CPS can be explained in terms of a system of parallel plate capacitors. Within the CPS, there is a movable diaphragm and a fixed substrate, each of which holds a capacitor plate. (See Figure 8 in Appendix D

for diagram.) The movable diaphragm faces the bronchial wall, with a plate on the inside surface. The plate on the fixed substrate lies parallel to the movable plate, separated by a thin space. When the airway constricts, pressure is applied to the stent. This pressure pushes the diaphragm plate and decreases the distance between the plates. As the pressure on the diaphragm increases, this causes an increase in capacitance, according to the inverse relationship between capacitance and distance between the plates ($C = \varepsilon A/d$)[17].

The pressure sensor itself is made of 16 deformable poly-silicon diaphragms that are positioned above insulated substrate electrodes. The substrate is composed of non-conducting fused silica material, which minimizes the interfering capacitances that could deter the accuracy of the measurements. Between the electrodes, a layer of dielectric insulation protects the CPS from damage at high pressures[83]. The CPS diaphragms, along with the rest of the electronic components in the stent, are coated in liquid crystal polymer (LCP), a biocompatible material.

When there is no abnormal constriction, we estimate that the resting pressure on the CPS will be approximately 1.09 bar, which corresponds to a resting capacitance of 5.98 pF(custom made to fit our sizing needs). For explanation of calculations within the CPS, see Appendix G1. We chose a pressure value of 1.14 bar (6.06 pF), corresponding to a 20% constriction of the diameter of the bronchus as our threshold at which the asthmatic is alerted of an attack. After further testing of our device, this threshold might vary. For more information on the thresholds, see Appendix AF.

Both the resting pressure (1.09 bar) and the threshold pressure (1.14 bar) fall within the CPS's pressure detection range of 0.3 bar to 1.3 bar[83]. Based on these calculations, we expect that the changes in capacitance between the baseline and threshold values will be significant enough to provide a reliable method of detection. The resolution of the CPS at 1 bar is +/- 0.05 mbar, and its absolute accuracy is +/- 1 mbar, which allows the CPS to accurately differentiate between pressures which only differ by .001 bar or less. Since our baseline and threshold pressures differ by .05 bar, this resolution will be more than sufficient for the changes we will detect. The temperature range in which this CPS works optimally is -40° Celsius to 85° Celsius, and body temperature (37° Celsius) is safely within this range[83].

For more information on the calculations behind these pressure and capacitance values, see Appendix G1. For more pressure and capacitance values associated with an asthma attack, see Appendix G1. The mechanism by which the patient is notified of this bronchoconstriction and capacitance change is detailed later in the paper.

C. External Notification and Powering Component

This implantable device will be accompanied by an external notification and powering component. This component consists of a purchased 6 cm x 6 cm patterned plate, which transmits power to an encapsulated receiver coil housed on the metallic strip of the stent and provides the power necessary for our device[39]. Additionally, a 6 cm x 6 cm x 1 cm box made of lightweight flexible biocompatible polymer is fused to this plate to contain electronics, including a receiver, vibration motor, switch, battery, and buzzer, as well as the circuitry connecting them. See Figures 9-11 in Appendix E for diagrams. The box containing these electrical components is hermetically sealed to avoid gaseous interference or moisture damage. The external device is placed on the outside of the upper torso and aligned with the internal device. The exact placement of the external device can be determined by the physician after a bronchial location for the internal device has been chosen. The interface is secured to the chest using a hydrocolloid adhesive patch (specifications in Appendix C). The adhesive lasts approximately 7 days when worn continuously, and must be replaced about once a week (#ref). The battery lifetime is approximately 6 years, after which it must be recharged using a simple wall-plug charger (See Appendix G3 for calculations). When the external device is notified of an attack, it can either vibrate or make a buzzer noise to notify the patient.

D. Electronics

a. Internal Circuitry

The circuitry in the electronics of the internal stent is built onto an application-specific integrated circuit (ASIC) which is custom-made by Texas Instruments using their 130-nm CMOS process[19]. This integrated circuit enables powering of our device through a custom-engineered capacitive power-storage circuit and interpretation and transmission of data through two circuits modeled after the ASIC detailed in the paper "Fully Wireless Implantable Cardiovascular Pressure Monitor Integrated with a Medical Stent"[19]. The power circuitry is discussed in the "Power" section of this paper.

In order for changes in capacitance to be interpreted, we must convert capacitance into time. We employ a capacitance-to-time circuit operating on the basis that as capacitance increases, the charge-up time of a capacitor also increases.

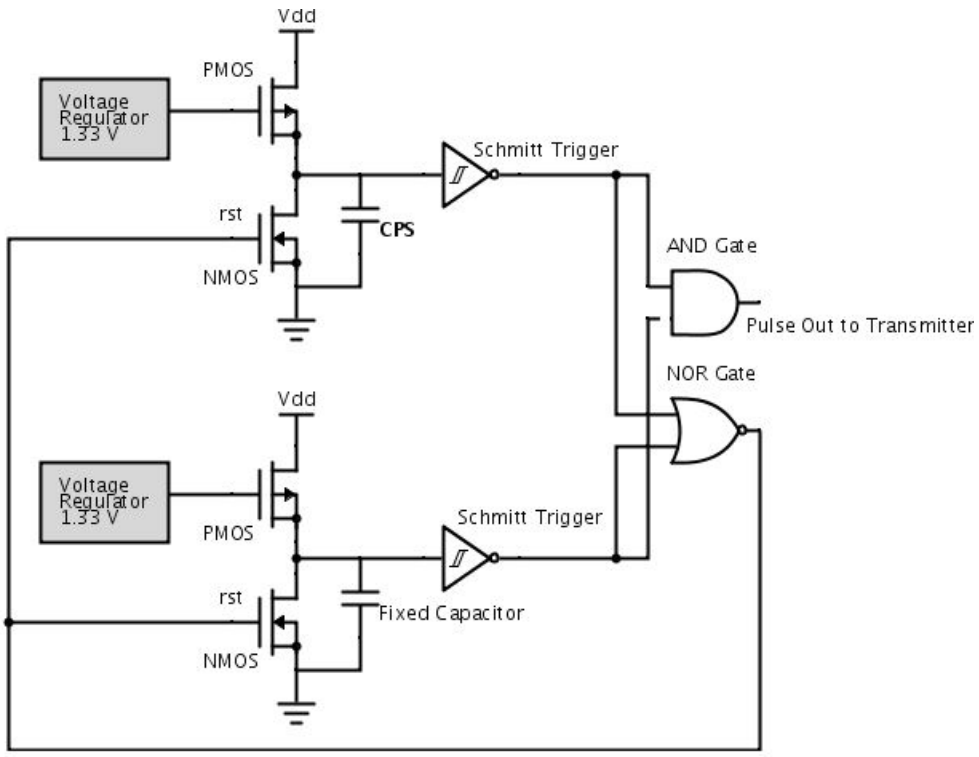

Figure 4: Capacitance-to-Time Circuit

Based on: Chow, Eric Y. "Simplified schematics of capacitive-to-time circuit." *Fully Wireless Implantable Cardiovascular Pressure Monitor Integrated with a Medical Stent*. 2010.

The above circuit (modified from the capacitance-to-time circuit in the paper) compares the charge-up time of its fixed capacitor to that of our variable CPS. The fixed capacitor is set to the capacitance corresponding to that of an asthma attack as defined earlier at 6.06 pF. For this purpose, we will use an Accu-P Thin Film 6.6 pF capacitor adapted to be 6.06 pF[1]. Therefore, whenever the constriction of the bronchus applies a pressure on the CPS resulting in a capacitance larger than 6.06 pF, the CPS has a larger charge-up time than that of the fixed capacitor. Our circuit is designed so that whenever this condition is true, a pulse is sent from the AND gate to the transmitter connected to this circuit. When the CPS has a capacitance of less than 6.06 pF, indicating safe bronchial pressures, no pulse is sent from the AND gate. This ensures that a signal is sent to the transmitter exclusively when the bronchial constriction is enough to be indicative of an asthma attack.

After each pressure measurement, the capacitor's charge must be reset to 0 Coulombs. We have deviated from the circuit outlined by and added a NOR gate wired to both Schmitt triggers and both pins labeled 'rst' in order to drain the capacitor and reset the circuit once both capacitors are fully charged[19]. For more details about the specific mechanisms of this circuit, including an explanation of the Schmitt triggers and logic gates please see Appendix G2.

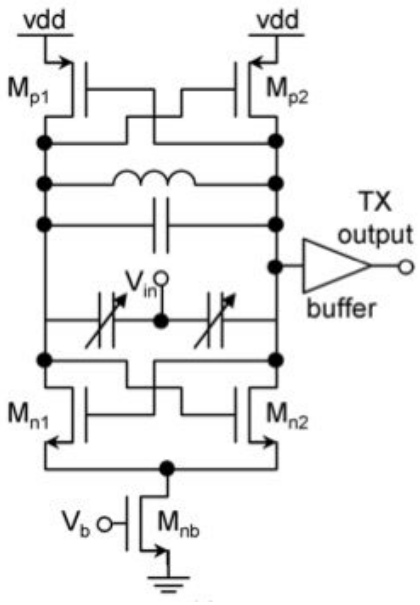

Figure 5: Transmitter Circuit

Chow, Eric Y. *Fully Wireless Implantable Cardiovascular Pressure Monitor Integrated with a Medical Stent*. 2010.

The transmission of the signal out of the body occurs through 2.4-GHz wireless transmitter built onto the ASIC[55]. The transmitter modulates and amplifies the incoming signal, sending it out at an RF frequency via the connected 3.2 mm x 1.6 mm x 0.6 mm ceramic miniature loop antenna[41]. This flat, rectangular antenna is oriented parallel to the receiver in the external component in order to achieve maximum signal strength and transmittance.

b. External Circuitry

The circuitry of the external component of our device is responsible for three functions: power regulation, conversion of the input signal from the transmitter to an alert, and monitoring battery levels. We have designed the following circuit based on the capabilities of a microcontroller to utilize timer functions as well as interpret and send programmable binary signals. (These functions can be programmed in C or Analog). The specific microcontroller we have chosen is the MSP430G2553IRHB32R microcontroller from Texas Instruments[63,66].

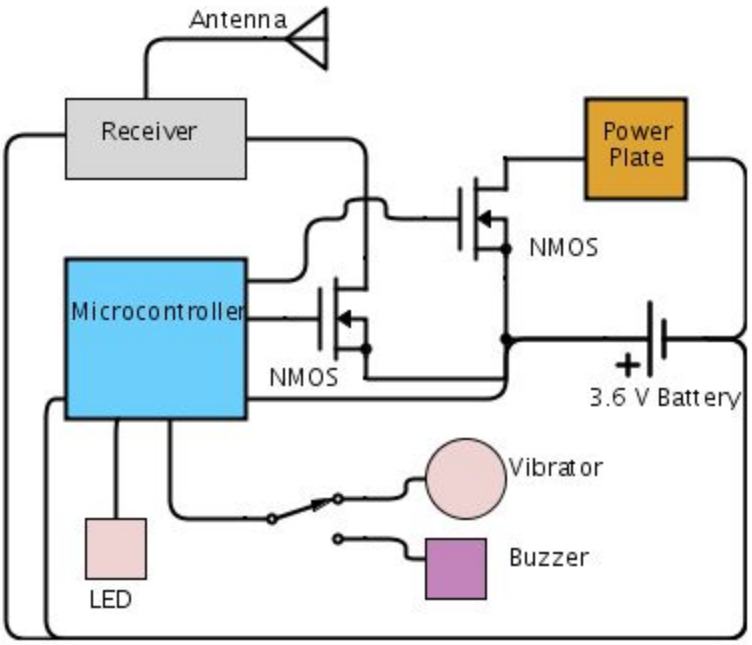

Figure 6: Circuitry of External Component

In order to perform these functions, the circuit shown above operates in 20-minute cycles carried out by the microcontroller's timer function; this process also conserves power. This does not affect the accuracy or precision of our device because, as previously mentioned, the onset of an asthma attack develops within 15-30 minutes after exposure to an allergen. Thus, having a device that only takes measurements every 20 minutes will be well within the time range of when an asthma attack develops.

The cycle begins when the microcontroller enters its "active" mode, which lasts 0.02 s, the minimum amount of the time the microcontroller requires to run its code. The microcontroller signals to the two transistors between the powering plate and the battery and between the 2.4 GHz receiver and the battery to allow for the passage of current, essentially switching "on" the power to the rest of the external component. This simultaneously allows the patterned plate on the external component to send power to the internal stent and the receiver to obtain transmissions that may result in an asthma alert. (Full details of power will be explained in "Power" section).

During this 0.02 s, the entirety of the implantable circuitry described in the previous section runs, and, if applicable, the receiver may obtain a transmission corresponding to an asthma attack. The receiver filters and demodulates the signal and sends it to the microcontroller, which, recognizing the signal as indicative of an attack, sends a burst of voltage to either the vibrator or buzzer to elicit a vibration or sound alert. This setting is customizable by manually adjusting an external switch on the box from vibrate to sound.

The microcontroller then enters “standby” mode for the following 20 minutes until the timer once again initiates its “active” mode.

The last function of our external circuit is to alert the patient when battery levels are low. The microcontroller is programmed, using the timer function, to check on the voltage levels every month and compare them to set threshold values. Once the voltage outputs of the battery falls below the threshold, the microcontroller outputs a voltage to a 5V Lumex LED that will continue to be lit until the battery is recharged by the patient.

c. Power

A rechargeable 3.6V battery of dimensions 5.8 cm x 3.6 cm x 0.47 cm within the external component powers our entire device. This 780 mAh battery acts as the power source for the power plate and the microcontroller. The power plate then provides power to the internal component (#ref).

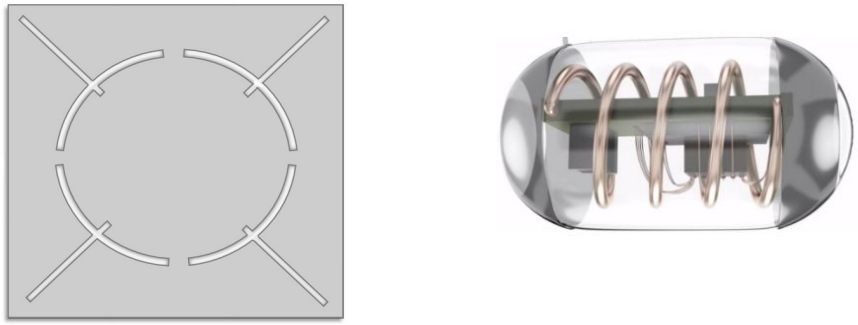

Figure 7: Power Plate and Microcapsule (not to scale)

Based on images from: Ho, John S. *“Wireless power transfer to deep-tissue microimplants*. 2014

We are purchasing power source technology detailed in the paper “Wireless power transfer to deep-tissue microimplants”[39]. This novel device employs a method termed “midfield powering” which utilizes far-reaching, decay-resistant waves. This methodology functions through the patterned plate emitting 1.6 GHz RF signals, generating sufficient power for our device while maintaining low energy input. The power source adheres to the patient’s chest, approximately 4 cm from the internal component and thus, delivers to it approximately 210 µW of power[39]. See Appendix G3 for details as to how these number were obtained.

A biocompatible epoxy-coated microcapsule (2 mm diameter, 3.5 mm height) located on the metal plate of the internal stent harnesses this energy to power the electronics within the body. A small copper coil within the microcapsule receives the midfield waves[39]. This microcapsule contains a rectifier circuit, which converts this AC current to the DC current that runs through the ASIC. We have

modified the circuit within the microcapsule to omit two parts that are unnecessary for our purposes. The first one is an LED later in the circuit that will be replaced with a simple diode, as we have no need for light inside the body. Secondly, the extra integrated circuit for pulse control will also be removed as we need continuous power from the microcapsule to the ASIC. The power instead goes directly to the electrodes connected to our ASIC and transmitter.

Block	Simulated Power Consumption	Measured Power Consumption
2.4 GHz Transmitter	1.394 mW	1.1506 mW
Voltage Regulator	32.02 μW	39.38 μW
Base Capacitance-to-Time Circuit	47.4 μW	58.30 μW
Sensor Capacitance-to-Time Circuit	53.15 μW	65.37 μW
System without Transmitter	64.86 μW	202.70 μW

Table 1: Power Consumption of ASIC

Adapted from: Chow, Eric Y. "Measured and simulated power consumption of ASIC components." *Mixed-Signal Integrated Circuits for Self-Contained Sub-Cubic Millimeter Biomedical Implants.* 2010.

This microcapsule only requires ~10 μW to power, leaving ~200 μW to power the rest of the system. As shown in table above, the ASIC takes 202 μW to power (as we do not include the FeRAM in our device) and would be easily powered by the microcapsule[20]. However, the transmitter itself requires energy magnitudes greater than that provided by our power source.

To account for this, we propose an energy storing capacitive circuit built into the beginning of the ASIC containing the capacitance-to-time circuit and transmitter. Since our ASIC will be custom-made, this circuit can easily be integrated. This circuit contains a 5.78pF capacitor modified to an appropriate size that stores enough charge that, when discharged, provides 1.4 mW; this is enough power to power the transmitter, which requires 1.35 mW for the transmittance of a single signal. The additional 0.05 mW accounts for any potential variations that might arise.

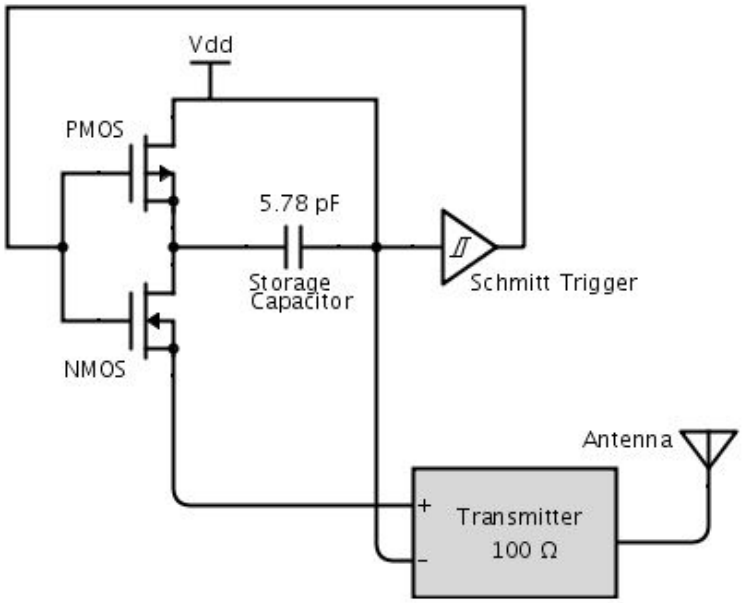

Figure 8: Power Storage Circuit

This circuit, pictured above, allows charge to stored in the 5.78 pF capacitor so it can be discharged when the transmitter needs to be powered. This circuit is regulated by a set of PMOS and NMOS transistors which, as previously explained, are electronic switches that allow charge to flow. These two switches will always be in opposite states of each other and will never simultaneously both be open or closed. When the capacitor is charging, the PMOS transistor will be 'open' and let charge run in to build up on the capacitor. When the capacitor has fully charged, the PMOS transistor will 'close' and the NMOS transistor will 'open' allowing for the built up charge to go directly to the transmitter. This circuit is also wired to the capacitance to time circuit so that all the electronics can be powered by this power source. This circuit will be charged for 0.02 s. This is more than sufficient for our capacitor to discharge and for a signal to be transmitted out of the body, all of which takes only 96.9 ns. Please refer to Appendix G3 for a full list of power calculations.

This devised method of powering will theoretically deliver enough power to our implanted component; however, this cannot be proven without further testing of our device. We foresee refinement and successful implementation of the power source during the prototype development phase of this project. In the development of our device design, we also considered RF wireless powering technology, which has been implemented with varying levels of success in comparable implantable bioMEMS technology. However, because the RF field is potentially harmful and delivers insufficient power to our internal component, the better proposed option was ultimately chosen. Please see Appendix G2 for more details on this process.

For a diagram detailing the connections between the all of circuits mentioned above, please see Appendix G2).

VII. Implementation of Device

A. Lifetime of Asthma ConStent

The life expectancy of the ConStent depends primarily on the biocompatibility of the device and the longevity of the implanted electronics. Based on existing research, we expect that the biocompatibility of the materials will prevent immune and other bodily responses which might otherwise impair the device's functionality. The polyurethane that covers the stent and the LCP that seals the electronics and the external component protect our device from granulation, water damage, and other environmental influences that could degrade our technology[68].

Lifetime is also affected by the battery life found in the external component. The battery is calculated to last for six years, but can be easily recharged at that time. See Appendix G3 for detailed calculations.

Without further testing of our electronics, we cannot calculate a definitive life expectancy, since that will depend on unforeseeable conditions within the bronchus, as well as random electronic malfunctioning. However, with proper maintenance (detailed below), we expect the ConStent to last approximately ten years, which is comparable to the life expectancy of other implanted devices[74].

B. Maintenance

To allow our external notification component to have the greatest lifetime, the patient must practice proper maintenance. The adhesive patch for the external portion must be replaced approximately every seven days to ensure the external component stays in place (#ref), which is vital for powering the implanted component. To guarantee that the stent has not migrated and the electronics are still secured, a doctor performs a yearly checkup using a simple bronchoscopy procedure. An additional advantage of our polyurethane and LCP seals is the ability to easily slide the stent from the bronchioles if removal is necessary[68].

C. Replacement

The stent portion of our device is designed to last for the span of a lifetime, assuming the patient uses proper maintenance as described above. However, the electronic components of our device might need to be replaced to ensure optimal functionality. The yearly bronchoscopy should indicate if your device is not working properly, and if so, it would need to be replaced immediately.

D. Cost to Manufacture and Implement the Device

The cost of the Asthma ConStent totals up to $539.34, which accounts for the one-time production cost of the implantable and external components; however, there is an additional, recurring cost of $2.52 per week for the necessary adhesives that totals to $787.14 over six years , for an overall cost of $1,326.48. (Appendix E2) However, we expect that we could decrease the cost of adhesives by manufacturing our own to better suit our device's needs; additionally, we would expect insurance to fully cover this recurring cost.

For a detailed list of manufacturer details and costs for each component, see Appendix C.The cost to implement the device also depends on testing and the timeline of development outlined in the last section of the paper.

VIII. Procedure and Implantation Cost

The insertion of the device must be conducted by an expert pulmonologist.. The procedure time for insertion of the device is approximately 30 minutes to 1 hour[38]. After being placed under local anesthesia (through IV) , the procedure begins with stent sizing when the doctor analyzes a patient's selected bronchial branch diameter using a flexible bronchoscope. To implant the stent, a flexible guide wire is pushed through the bronchoscope to a point past the location of desired stent placement. Then a constrained stent compressed to half its diameter is advanced over the guidewire. The stent is positioned so that the strip of electronics is parallel to the power source. Once the stent is in the desired position, it is deployed by the delivery catheter. The overall bronchoscopy cost is a total fair price of $1,946[109].

IX. Strengths and Weaknesses

A. Weaknesses

The cost of our device itself is less than $100 over our target goal of $500, but the recurring costs of adhesives add up to make this device more costly than we had initially hoped. However, this is still a reasonable cost when compared to the cost of hospital stays and missed work days which can result from severe asthma attacks.

In regards to the stent, a major weakness is the harmful effects that may arise in the bronchi, such as granulation or inflammation that could cause discomfort in the patient[68]. The stent may migrate out of proper placement, which could cause an infection[34]. However, our coating and antimigration methods should help combat these problems. Moreover, mortality due to stent placement is rare (#ref).

Another weakness is that the patient cannot tell whether the implantable electronics in our device is functioning or not. Although our electronics could theoretically last throughout a patient's lifetime, it is likely that the electronics may stop functioning before this time. To account for this, we have recommended yearly check-ups with a pulmonologist.

Additionally, our device transmits RF signals through the body from the implanted component to the external component, and some research indicates that the effects of internal RF signaling on human tissues may be detrimental[19].

B. Strengths

Our chosen design meets or approaches most of our target goals and falls short in the cost criteria. Our device requires only minimal patient involvement: the detection method is fully automated, and the battery requires charging very infrequently. Our device would notify the patient of an asthma attack approximately 20 minutes after allergen exposure, which matches our target goal. The procedure for implanting our device would take approximately 30 minutes-1 hour and would require local anesthesia with no stitches or hospital stay required[38], therefore meeting our target goal. The durability of our device meets our target goal and lasts approximately 10 years, assuming the patient practices proper maintenance.

Table 2: Comparison of Asthma ConStent to Target Goals

Criteria	**Target Goals**	**Chosen Design**
Level of Patient Involvement	Fully automated and requiring no patient involvement	Fully automated, but requires weekly maintenance
Notification Time	20 minutes or less after allergen exposure	20 minutes after allergen exposure
Complexity of Implantation	Procedure time 30 minutes or less with local anesthesia and no stitches required; outpatient procedure	Procedure time 30 minutes-1 hour with local or general anesthesia and no stitches required; outpatient procedure
Durability	No periodic maintenance for internal implant Lasts > 5 years	Lasts ~10 years
Cost	Less than $500	~$550, with recurring cost of $2.50/week

Other strengths of our device include its ability to handle natural human variation. The ConStent could be used for a wide range of adults with allergy-induced asthma, because only the size of our stent would differ between individuals. Stents are a common medical device already produced in various sizes and materials, so it would be feasible to manufacture our device in different sizes Additionally, our external component can be placed in various positions on the torso. This gives the pulmonologist more options of where to place the internal device within the lobar bronchi, allowing for optimal comfort in the patient.

Our chosen pressure sensor can measure a wide range of pressures with high sensitivity. Therefore, our device will be able to detect relevant changes in pressure even if the tested values of the device differ significantly from our predicted values.

The circuitry requires minimal power and performs the desired tasks quickly so that the battery only requires recharge every 6 years. The powering is done wirelessly and with midfield waves, which have minimal health risks to the patient.

X. Proposed Timeline of Development

The timeline for the development of a new medical device depends primarily on the process required by the FDA to test and approve the device. Our device would be classified as the most heavily regulated Class III device. Therefore, we need a Premarket Approval (PMA) from the FDA before it could be produced and marketed[80]. The application for a Premarket Approval requires a detailed description of the device, as well as a summary of the studies performed with the device, and convincing evidence about the device's safety and efficacy[79].

Before we could apply for a PMA for our device, we would perform preliminary laboratory tests, such as using machines to simulate bronchial constriction and using in vitro studies to confirm the biocompatibility of our materials. Next, we would move into animal testing; This preclinical stage would probably take about 2-3 years[43].

To start clinical testing, we would first need to receive an Investigational Device Exemption (IDE) from the FDA, and an approval from the local Institutional Review board, which would take 6 to 12 months. During this time we would partner with various clinical sites to implement our studies. Then, we would do clinical trials on less than 100 participantsbefore expanding our studies to the pivotal phase that involve over 1000 patients We would monitor this over the course of 2-3 years in order to monitor the safety and efficacy of our device[43].

The whole process of clinical trials would take at least six years. After completion of these trials, we would be able to apply for a PMA and implement our device. We expect that the entire process from lab testing to PMA would take about ten years.

XI. Conclusion

Our goal was to develop a device to detect the onset of an asthma attack in patients with severe asthma. We have successfully designed an implantable pressure sensor that detects bronchial constriction, which is a defining feature of an asthma attack. Upon early detection, the implantable device sends a signal to an external component, which alerts the patient so that he or she is able to take the necessary medications before the attack becomes too severe. Our device provides a sound basis for a new technology which can realize the management of severe asthma.