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IMPLEMENTING VENTILATION SYSTEM FOR PREMATURE INFANTS BY USING IOT

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ABSTRACT

This system design of a premature infants monitoring system based on wireless technology. A prototype is developed which gives a reliable and efficient baby monitoring system that can play a significant role in higher health care kid. In the baby monitoring system, the necessary parameters of the infant like, gas sensor, ventilation system, and GSM module. Baby lungs use the reverse pressure generated by contraction motion of the diaphragm to suck in air for breathing. A contradictory motion is used by a ventilator to inflate the lungs by pumping type motion. A ventilator mechanism must be able to deliver in the range of 10 – 30 breaths per minute, with the ability to adjust rising increments in sets of 2. Along with this the ventilator must have the ability to adjust the air volume pushed into lungs in each breath. The last but now the least is the setting to adjust the time duration for inhalation to exhalation ratio. Apart from this the ventilator must be able to monitor the baby's blood oxygen level and exhaled lung pressure to avoid over/under air pressure simultaneously. GSM module is designed for wireless radiation monitoring through Short Messaging Service (SMS) will be displayed in the mobile through notification. If the readings seem abnormal, the caretaker along with the parents will get an alert message. This monitoring system is a highly efficient IOT based system for real-time monitoring with the best security measures and alerts.

Keywords: GSM module, Risk factors, infectious organisms and exhaled lung pressure.

1. INTRODUCTION

Respiratory disorders or lung diseases are conditions such as asthma, lung cancer, cystic fibrosis, pulmonary hypertension, tuberculosis (TB), emphysema, mesothelioma, to name a few [20]. These are either inherited genetically or are caused by the long-term exposure to external irritants that damage the lungs and the airways. Respiratory diseases produce health complications and life-threatening conditions if they are left untreated [20]. They make up to five out of the thirty most common causes of death around the world: Chronic Obstructive Pulmonary Disease (COPD), lower respiratory tract infection, lung cancer, TB, and asthma are ranked third, fourth, sixth, twelfth, and twenty-eighth, respectively [1]. More than one billion people around the world suffer from acute or chronic respiratory conditions [1]. Respiratory disorders are the third leading cause of death in Canada and the United States both in adults and infants [2-3]. In Canada, one in five people have a respiratory disorder. Over two million Canadians suffer from asthma, one of the leading causes of hospital admissions among children [4]. In 2014, lung cancer caused more cancer deaths among Canadians than colorectal, breast and prostate cancer combined [5]. The leading cause of hospitalization among adults is COPD, which also accounts for more than 10% of all disability-adjusted life-years (DALYs). DALYs is a metric that estimates the amount of active and productive life lost due to a condition [8]. Respiratory diseases impose a significant burden on the Canadian economy, particularly COPD, asthma, and lung cancer. Respiratory disorders account for 6% of annual healthcare costs in Canada. In 2014, respiratory disorders were estimated to cost the Canadian economy over $12 billion every year according to an analysis by the Conference Board of Canada [8]. The internal organs that are most vulnerable to injury and infection from the external environment are the lungs. The constant exposure to chemicals, such as toxic smoke of biomass fuel, particles, inhaling polluted outdoor air, and infectious organisms in the air cause lung infection. By 2030, it is estimated that 12 million Canadians will be affected annually by respiratory disorders [6] if there are no further enhancements and strategies made for dealing with respiratory diseases. The annual economic burden is also projected to be double by 2030. Innovative strategies, policies to further reduce and modify the risk factors and treatments must be developed in order to reduce the imminent burden of these conditions on the economy and the health care systems. A potential area of research is in the development of technologies to assist babies with respiratory disorders.

2.LITERATURE REVIEW

BREATHING ANATOMY

Breathing is essential for survival. The human body can live without food for 3 weeks and water for 3 days, but only 3 minutes without air. Normal bodily functions cease to occur when brain is starved of oxygen. The act of breathing is called pulmonary ventilation [1] and is described as the process of air flow into and out of the lungs from the atmosphere during inspiration (breathing in air) and expiration (breathing out air) [11]. The air movements inside the lungs are governed by the principles of the gas laws. Pulmonary ventilation is dependent on three types of pressure [14]: atmospheric pressure (P_{atm}); intrapleural pressure (P_{ip}), the pressure within the pleural cavity; and intra-alveolar pressure (P_{aiv}), the pressure within the alveoli. The air flows into the lungs due to the difference in the pressure. The air flows down a pressure gradient from an area of higher pressure to an area of lower pressure. Atmospheric pressure is greater than intra-alveolar pressure and intra-alveolar pressure is greater than intrapleural pressure [12]. The same principle applies during expiration, when air flows out of the lungs. During exhalation, pressure within lungs becomes greater than the atmospheric pressure.

MECHANISM OF BREATHING

The two major steps involved during respiration are inspiration and expiration. When air enters the lungs, the process is called inspiration, and when the air leaves the lungs, it is called expiration. This process is shown in Figure 2.1. One full sequence of expiration and inspiration is called a respiratory cycle. The two general muscle groups that are used during normal inspiration are the diaphragm and the external intercostal muscles [9]. When a person takes bigger breath, additional muscles are required. The diaphragm moves towards the abdominal cavity when it contracts, this creates a larger thoracic cavity and hence there is more space for the lungs. The rib cage then expands and the volume of the thoracic cavity increases due to the contraction of the external intercostal muscles [19]. Ribs move upward and outward due to the contraction of the external intercostal muscles. This increase in volume leads to a decrease in intra-alveolar pressure [1]. Hence, a pressure lower than atmospheric pressure is created. This creates a pressure gradient that then drives air into the lungs. The expansion and contraction of the thoracic cavity causes inspiration and expiration respectively. The intra-alveolar (i.e., the pressure within the alveoli) and intrapleural pressures (i.e. the pressure within the pleural cavity) are dependent on certain physical features of the lung [19]. However, the ability to breathe i.e., to have air enter and leave the lungs during inspiration and expiration, respectively, is dependent on the air pressure of the atmosphere and the air pressure within the lungs [11].

RESPIRATORY MECHANICS

During normal inspiration, negative intrapleural pressure is generated. In other words, a pressure gradient is created between the atmosphere and alveoli, resulting in airflow. In mechanical ventilation, however, the pressure gradient results from increased (positive) pressure of the air source. At the airway opening, peak airway pressure (Pa_o) is measured. It is the total pressure needed to push a volume of gas into the lungs [14]. This pressure is composed of the elastic recoil of the lung and chest wall (elastic pressure), the inspiratory flow resistance (resistive pressure) and the alveolar pressure present at the beginning of the breath (positive end-expiratory pressure [PEEP]) as seen in Figure 2.2. Hence, peak airway pressure is equal to the resistive pressure plus the elastic pressure plus the PEEP [15]. End-expiratory pressure in the alveoli is normally the same as atmospheric pressure. When alveoli fail to empty completely, the end-expiratory pressure may be positive relative to atmosphere. This pressure is called intrinsic PEEP or Auto PEEP. The end-inspiratory pressure represents the elastic pressure once PEEP is subtracted. The difference between the peak and the plateau pressure is the resistive pressure. The main factors that affect breathing are the levels of oxygen and carbon dioxide in the blood, the blood's pH level, and any respiratory disorders such as COPD, asthma, lung infection or collapsed lung.

MECHANICAL BABY VENTILATORS

A mechanical baby ventilator is a device that aids a baby to breathe when they are having difficulties; for example, if they are recovering from a surgery or serious sickness or have difficulty breathing on their own for any reason (e.g. a critical illness). When using a baby ventilator, a hollow tube or a mask is placed in the baby's mouth to connect them to the baby ventilator. Airflow is then pushed into baby's' lungs via the mechanical baby ventilator to help them breathe. Babys remain on the baby ventilator until their condition is improved or until they can breathe on their own. The two types of mechanical ventilation include the following: Invasive ventilation: usually performed in the intensive care unit with a tube inserted into the baby's airway. Noninvasive ventilation: usually a mask that goes around a person's mouth; it can be used at home by people that are facing respiratory difficulties.

3. PROPOSED METHOD

The ventilator we here design and develop using arduino encompasses all these requirements to develop a reliable yet affordable ventilator to help in times of pandemic. We here use a silicon ventilator bag coupled driven by servo motor with one side push mechanism to push the ventilator bag. Our system makes use of blood oxygen sensor along with sensitive heart Beat sensor to monitor the necessary vitals of the baby and display on a webpage using IoT. To adjust the time duration for inhalation the option command given in the IoT application to set. The entire system is driven by arduino controller to achieve desired results and to assist baby incubator. Health data's such as Temperature, Heart rate and SPO2 collected by an inter-body from wearable physiologic sensor is hub using SBC. Daily activities such as position, sound and moisture detection. The Wearable Hub can transmit the data through Smart band when the person wears the band in the receiving system enabling human Machine interaction through SBC.

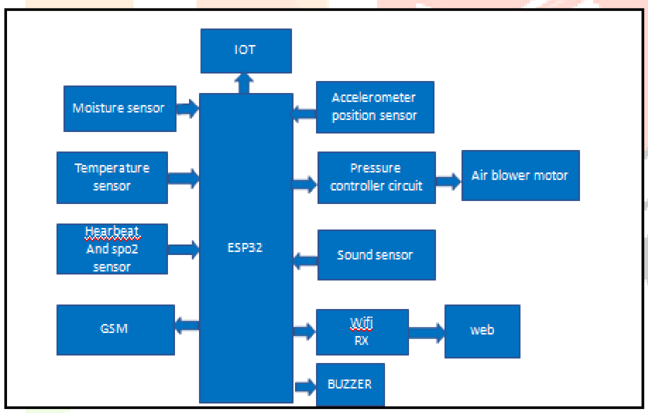


FIG: 1 BLOCK DIAGRAM OF PHYSICAL IMPLEMENTATION.

4. RESULTS AND DISCUSSION

This chapter describes the tests that were conducted in the prototype device after the flow sensor and alert codes were added to the microcontroller. The flow data were collected when the flow sensor was attached to the baby port of the PCM. The flow sensor reads real-time flow data. These flow data are important because they determine the airflow at the baby port of the PCM. The flow sensor was connected to the arduino, which monitored the baby ventilator performance parameters such as pressure, load, current, voltage and airflow. Flow data were collected every 30 ms. The PCM and a Quick Lung breathing simulator (test lung) were connected, by connecting a breathing circuit with a flow meter between them.

4.1 MOTIVATION AND OBJECTIVE OF THE EXPERIMENT

The motivation for doing the experiments is to make sure that the baby ventilator performs under different lungs and baby ventilator parameters. It is also to ensure that the baby ventilator is safe to use. The objective of conducting the experiment is to evaluate and validate the sensor used for the device under different baby ventilator and lung settings. A micro-controller device was programmed so that the controller of the system works as per the user's input data. Various experiments in the device with different settings were conducted to analyze and examine the output data generated based on the input provided, in order to check the efficiency of the system and the accuracy of the sensor used. The objective was to check whether the input volume of air is the same as the output volume of air at the PCM port before it goes to the test lungs and whether the sensor is collecting data for the desired duration of a breath cycle. Testing the added safety alarms for various conditions is also one of the objectives of the experiments. Various tests were done to gather more information, which could help the clinicians further assess the baby's condition.

4.2 TIDAL VOLUME TEST

General requirements from ISO 80601-2-79:2018 (Medical Electrical Equipment) specify that for volume-controlled breath types, during the testing, the error of the delivered volume of individual breaths shall not deviate by more than 35%, and that the delivered volume averaged over a one-minute interval shall not deviate by more than 25%. In order to conduct the tidal volume test, a flow meter was connected to the baby port of the PCM in the BABY VENTILATOR as shown in Figure 4.1. The test lung and the baby ventilator device are connected by a breathing circuit. When the baby ventilator is switched on, the PCM is squeezed by the arm, up to a certain level, according to the input baby ventilator parameters. When pressure the PCM, the air passes from the PCM to the baby port. The flow sensor is attached to the baby port. The air flows through the flow sensor and the flow rate is detected by the sensor. This flow rate data is recorded to further evaluate the performance of the baby ventilator device. The pressure values are also being recorded simultaneously along with flow rate values. The pressure during inspiration increases as the PCM squeezes and reaches a peak inspiratory value. The pressure then decreases during expiration. Tidal volume (V_t) is a baby ventilator input setting, which determines the volume of gas delivered by the mechanical baby ventilator. The tidal volume input must correlate to the actual volume delivered to ensure that the tidal volume requirements are accurately met. The volume of gas delivered by the mechanical baby ventilator to the lungs is dependent on the compression mechanism on the PCM, the compliance and resistance of the lung, the I: E ratio limits and the torque of the motor providing the mechanism movement.

4.3 METHODS

Once, the input tidal volume is set, the device runs for 1–5 cycles. As the PCM is compressed and released, the air that comes out of the PCM passes through the flow sensor through the breathing circuit and goes to the test lung. The flow data from the flow sensor are then recorded. The instantaneous peak inspiratory and expiratory pressures are displayed in the 20×4 display of the device. The pressure observed at every point is also recorded. Baby ventilator parameters, such as V_t and BPM, were varied, test lung parameters such as resistance were adjusted, and flow and pressure readings were recorded.

The data collected from the flow sensor were used to compute output tidal volume. The inspiratory and expiratory pressure, output volume and error between input and output volume for various test the bar graph for the input Vt and output Vt obtained from the flow data for different test procedures, which makes the comparison easier. The blue bar and orange bar represent input and output tidal volume, respectively. The output Vt is calculated by taking the integration of the set of flow data for each condition using MATLAB (MATLAB codes used for calculation are presented in Appendix B). The integration is done using the trapezoidal rule by approximating the region under the flow graph and calculating its area. The error percentage is calculated taking the percentage difference between the input and output tidal volumes. From the table, it can be observed that the error percentage for baby ventilator performance for 5 (cmH_2O/ (l/s)) and 10 (cmH_2O/ (l/s)) resistance is less than 15%. However, for higher resistance 20 (cmH_2O/ (l/s)), the error percentage is more than 15%. This is because by applying different lung resistance values, the lung volume at the end of the exhalation is increased. In presence of increased airway resistance, an increased transpulmonary pressure (the difference between the alveolar pressure and the intrapleural pressure in the pleural cavity) is required to produce a given tidal volume, and therefore the work of breathing is increased. Different control parameters can also create issues when they are at their maximum values.

Test Procedure 1: Compliance: 20 (ml/cmH_2O), Resistance: 20 (cmH_2O /(l/s)), PEEP: 10 cmH_2O, I: E ratio: 1:2

BPM	Input Tidal Volume (V_t) (ml)	Inspiration Pressure cmH_2O	Expiration Pressure cmH_2O	Output V (ml)	Error %
12	300	21.6	14.0	260	13.33
	400	30	13.9	370	7.50
	500	36.1	14.3	470	6.00
	600	39.1	13.5	550	8.33
14	300	23.6	14.7	270	10.00
	400	29.4	13.9	380	5.00
	500	35.2	41.5	480	4.00
	600	42.0	13.3	550	8.33
16	300	25.8	13.7	280	6.67
	400	33.4	14.3	390	2.50
	500	41.5	42.6	480	4.00
	600	44.8	13.6	560	6.67
18	300	26.5	14.0	280	6.67
	400	34.1	13.3	390	2.50
	500	42.6	13	490	2.00
	600	47.3	13.2	580	3.33

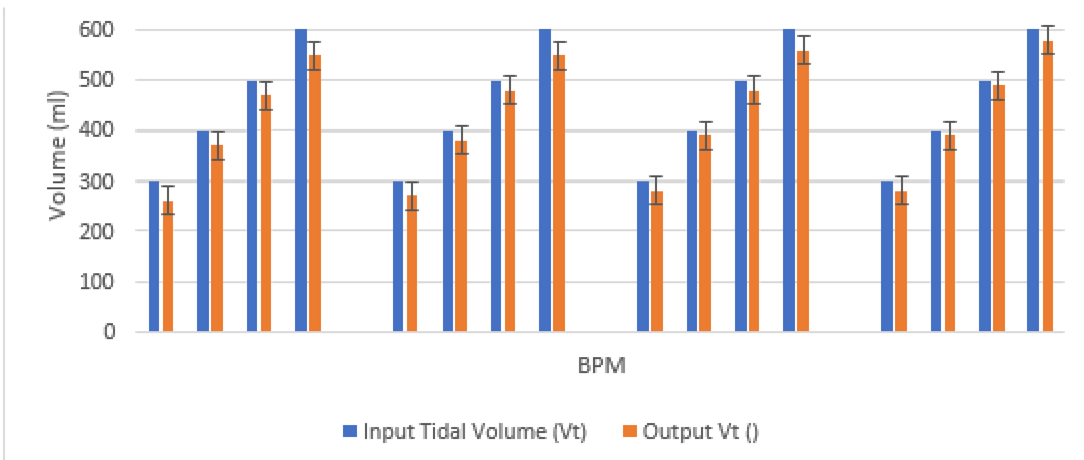
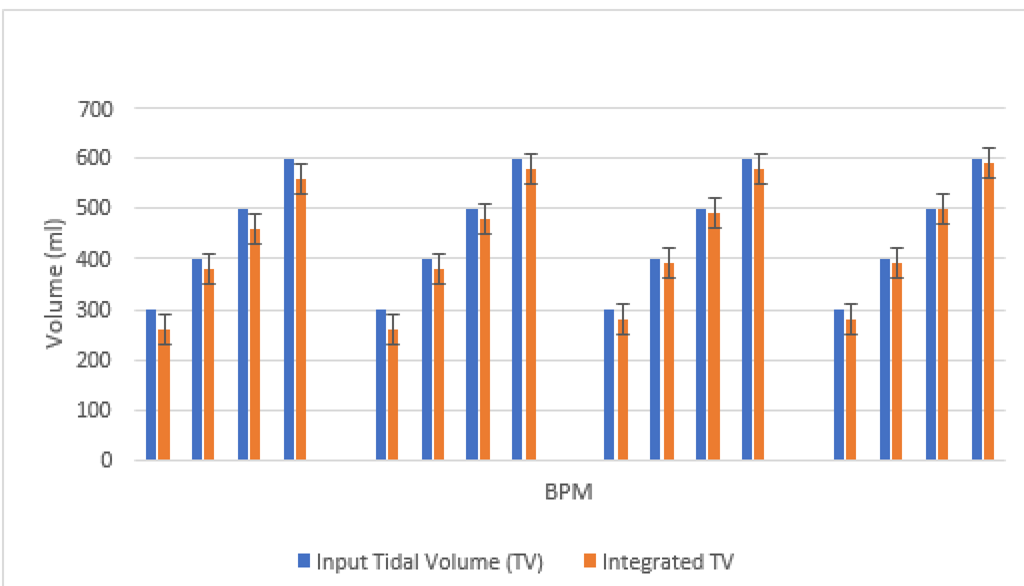


FIGURE 3: INPUT V_t AND OUTPUT V_t FOR TEST PROCEDURE 1.

Test Procedure 2: Compliance: 20 (ml/cmH$_2$O), Resistance: 5 (cmH$_2$O /(l/s)), PEEP: 10 cmH$_2$O, I: E ratio: 1:2

BPM	Input Tidal Volume (V_t)	Inspiration Pressure cmH$_2$O	Expiration Pressure cmH$_2$O	Output V (ml)	Error %
12	300	21.9	11.9	260	13.33
	400	27.3	12.4	380	5.00
	500	31.7	12.1	460	8.00
	600	35.2	12.4	560	6.67
14	300	21.7	12.0	260	13.33
	400	27.5	12.4	380	5.00
	500	32.4	12.0	480	4.00
	600	36.6	12.3	580	3.33
16	300	21.7	12.2	280	6.67
	400	28.3	12.4	390	2.50
	500	33.6	35.7	490	2.00
	600	37.8	12.5	580	3.33
18	300	23.2	12.4	280	6.67
	400	29.4	12.2	390	2.50
	500	35.7	12.5	500	0.00
	600	39.1	12.5	590	1.67

FIG 4: RECORDED DATA FOR DIFFERENT LUNG SETTINGS



Test Procedure 3: Compliance: 20 (ml/cmH₂O), Resistance: 50 (cmH₂O /(l/s)), PEEP: 10 cmH₂O, I: E ratio: 1:2

BPM	Input Tidal Volume (V_t)	Inspiration Pressure cmH_2O	Expiration Pressure cmH_2O	Output V (ml)	Error %
12	300	29.9	15.0	250	16.67
	400	41.3	14.2	340	15.00
	500	54.4	14.3	400	20.00
	600	62.6	15.1	450	25.00
14	300	33.7	14.9	250	16.67
	400	48.2	14.7	350	12.50
	500	60.2	15.1	400	20.00
	600	71.1	15.2	480	20.00
	300	36.1	14.7	250	16.67

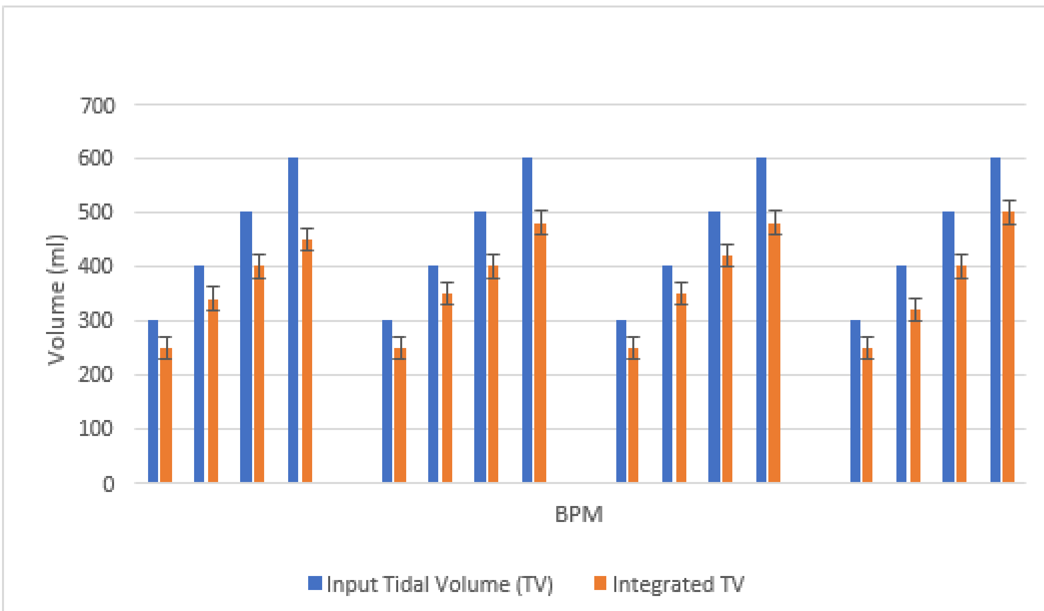


FIGURE 4: INPUT V_t AND OUTPUT V_t FOR TEST PROCEDURE 2.

In all the results obtained while testing with different baby ventilator and lung parameters, the error percentage between set or input tidal volume and output volume is less than 35%, which fulfils the condition from the general requirements from ISO 80601-2-79:2018 (Medical Electrical Equipment) Part 2-79: Particular requirements for basic safety and essential performance of baby ventilatory support equipment for baby ventilatory impairment.

5. CONCLUSION

The work presented in this thesis is towards the development of an intelligent and smarter controller module for the BABY VENTILATOR (Baby ventilator). A literature review was performed to understand the working principle of the baby ventilator device, and to identify the different types of existing low-cost open-source baby ventilators designs. A working prototype was developed for the controller module, and various sensors that were considered for the design were discussed and added. The prototype has been tested on the test lungs and the results. The baby ventilator operates effectively even with the combination of different baby ventilator and lung parameters. The flow sensor provided a fairly consistent result for delivered volume with an average deviation of 15%. The inspiration and expiration pressure is measured at all times and the pressure pattern can be used by the clinicians to evaluate a baby's condition. Safety alarms were tested for four conditions, and it worked competently. The current BABY VENTILATOR prototype that was worked on during this thesis includes user-controlled breath rate and tidal volume. It is a volume control device, and it has a number of alerts to make the system smart. A flow sensor was added to measure the flow rate at the output of the baby ventilator. This is important because knowing the flow rate ensures the correct amount of air that is being pushed into the lungs. Low volume and high volume of air in the lungs can both damage the lungs and put the baby's life at risk. Hence, adding a flow sensor and observing the output data makes sure that the volume of air provided in the lungs is adequate [16]. The on-going pandemic was a huge motivation for the start of this project. The pandemic resulted in limited healthcare resources. The risk of limited baby ventilators was one of the greatest concerns. Baby ventilator shortage affected every country around the world but mostly Low- and Middle-Income Countries (LMIC). A simple easy to build low-cost baby ventilator device such as BABY VENTILATOR, which can be made with readily available components and can provide a reliable baby ventilator solution, can be used in case of surge demand [11]. The relative low cost of this project could potentially provide a baby ventilator solution in other resource constricted environments as well [11].

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