

A Pressure Signal Apnea Monitor

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Introduction:

The thought process that respiratory depression, also known as hypoventilation, should be detected before the incidence of apnea has yet to be considered necessary. The current standard of care is similar to the process used by fire alarms. A fire is started, an alarm is sounded and the fire department arrives to put out the fire and minimize the damage. Yet, damage still occurs. The use of capnometry and pulse oximetry for the detection of hypoventilation performs in the same manner. A patient is struggling to breathe; s/he eventually stops breathing, the pulse oximeter continues to average the measurements and finally sounds an alarm. The capnometer continues to register CO₂ during the struggling breaths; eventually, it too sounds an alarm. The nursing staff is attending another patient and comes as soon as possible. 90 seconds later the staff arrives and the patient has stopped breathing, developed abnormal cardiac rhythms, and is developing respiratory acidosis. In the last couple of minutes the situation turned from one of prevention to one of crisis. Unfortunately the damage has been done and the nursing staff can only do their best to prevent further damage.

Background:

Two thirds of unexpected hospital deaths are attributed to hypoventilation [1]. After surgery, 0.5% to 5.2% of all patients receive

analgesics to control pain [2,3,4]. When analgesics are given in concentrations high enough to control pain, the drugs depress the patient's respiratory system. This causes the patient's respiratory rate (RR) to fall and the tidal volume (TV) of each breath decreases. As the TV decreases, each breath becomes inadequate to inhale sufficient oxygen. Supplemental oxygen is frequently delivered to patients post-anesthetic surgery. It is necessary to maintain adequate SpO₂ concentrations.

If supplemental oxygen is not successfully delivered to these patients, hypoxemia and or death may occur. Deaths caused by respiratory depression are preventable if detection is provided early on and supplemental oxygen is delivered correctly.

Apnea occurs in different forms. Central apnea occurs when the patient makes little to no effort to breathe. It is often caused as an effect of receiving too much of an opioid agent. Obstructive apnea occurs when the airway tissues are relaxed that they block the flow of gas, and thereby, reduce the volume of air that patient receives. Obstructive apnea occurs frequently during sedation by receiving too much of a sedative agent.

The current standard of care to monitor respiration is provided by either pulse oximetry or capnography [5]. Pulse oximetry measures the SpO₂ (saturation of peripheral oxygen) in the patient's arterial

blood. Pulse Oximeters continually average the SpO₂ and therefore are slow to respond to apnea. Furthermore when a patient's respiration is depressed and supplemental oxygen is being delivered, the O₂ concentration in his/her lungs remain relatively normal. Yet, during this period of hypoventilation the concentration of carbon dioxide (CO₂) in the patient's arterial blood rises (hypercarbia), and the patient becomes increasingly susceptible to abnormal cardiac rhythms. This can directly lead to cardiac arrest and or death. These patients become hypercarbic long before the pulse oximeter alarms [6,7,8].

Capnometry also has limitations that make it unsuitable for the detection of respiratory depression. Capnometers are unreliable when used with patients that are breathing spontaneously and when receiving supplemental O₂ [9,10]. The O₂ dilutes the expired CO₂ sample and therefore interferes with the measurement. Capnometry's sensitivity for the detection of apnea in non-intubated patients is 62%. This is low and due to this, capnometers miss more events of apnea than is tolerable for adequate patient care [11]. Finally, capnometers require a complex gas sampling system and infrared gas analyzer. This makes them more complex and expensive than a system based from a pressure transducer.

Significance:

Using a pressure signal to detect hypoventilation and apnea was first described by Guyatt et al. [12]. A pressure transducer connected to a nasal oxygen cannula is capable of detecting pressure fluctuations of approximately 1 cm H₂O during breathing. The cannula uses the nasal prongs as pitot tubes. A square root transformation of the original pressure

signal gives a linear relationship when compared to airflow. Measurements of pressure are accurate at high ventilation rates when supplemental oxygen was also given through the cannula [13,14,15,16,17,18]. Ballester et al. found that using nasal pressure has a respiratory detection rate of 96.8% [19].

Measuring respiratory pressure is the preferred method for the diagnosing of obstructive sleep apnea. However, this requires the nasal cannula to be accurately positioned at all times. Currently this is difficult for the medical staff to determine. Nasal cannula prongs frequently dislodge from their ideal location within the nares of the nose. Without the constant assistance of a nurse to reposition the cannula, the system would cause too many false alarms. This would only increase to the constant "alarm fatigue" that is experienced by all medical staffs. Therefore, we propose to develop a method to determine the position of the nasal prongs by analyzing the pressure changes caused by oxygen flow during respiration. This would provide the ability to identify a misplaced or dislodged cannula. This will increase the effectiveness of delivered supplemental oxygen and improve upon the current methods to monitor respiratory depression post-surgery.

Innovation:

New pressure transducer technology provides the ability to increase the accuracy of flow measurements. Using these measurements the pressure apnea monitor can detect when the cannula is inadequately positioned or dislodged. This is possible by measuring both the supplemental O₂ flow rate and inside of the nasal cavity. Using both we can calculate a

relationship between the backpressure within the nasal cavity as the O₂ fills it and the placement of the cannula. Thereby, providing a factor to determine when the cannula is misplaced.

This new development also enables us to detect airway obstructive and central respiratory depression. With obstructive respiratory depression the backpressure measured within the nasal cavity will fluctuate and contain higher frequency components during partial obstruction and subsequent snoring. Central respiratory depression will be identified by observing small and infrequent pressure swings that occur when the breath rate is slow and or the tidal volumes are small.

Additionally, this new technology provides for enhanced accuracy and efficiency in detecting pressure measurements. Our pressure transducer will use a second compensating solid-state diaphragm that automatically adjusts for the effects of gravity and temperature. This provides a system that is less affected by positional changes and temperature drift. We will process the pressure signal with a precision adjustable gain differential amplifier and a high-resolution analog to digital converter system to allow for precise and accurate measurements at very low pressures.

Preliminary Studies:

Figure 1 is a block diagram of the current prototype. It includes a traditional oxygen flow meter (Figure 2) and an inexpensive differential pressure sensor (MPXV5004DP,



Figure 2: Drawing of a traditional oxygen flow meter.

FreeScale Semiconductor, Austin TX). Both are connected to a divided nasal cannula (Figure 3).

The nasal cannula was placed on a mannequin that simulates an adult nasal cavity, oral cavity, larynx, and trachea. The simulated trachea was connected to a mechanical test lung (Figure 6). Preliminary data (Figure 4) shows the pressure measured in the right prong of the nasal

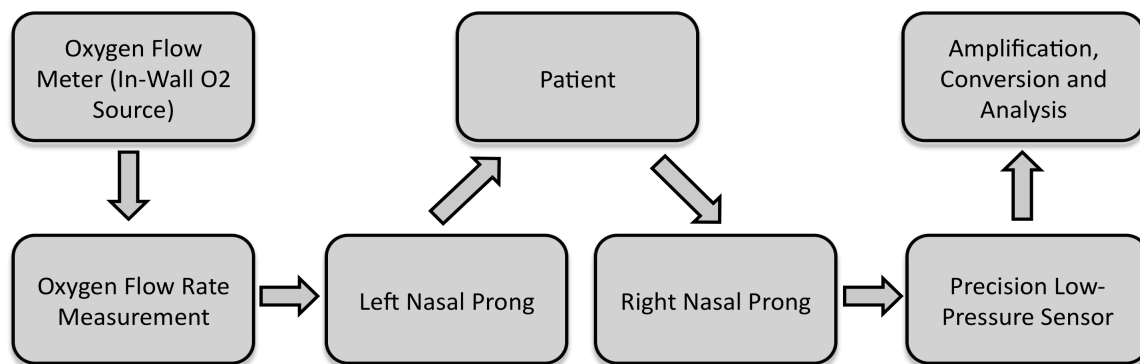


Figure 1: Block diagram of the pressure signal apnea monitor prototype.

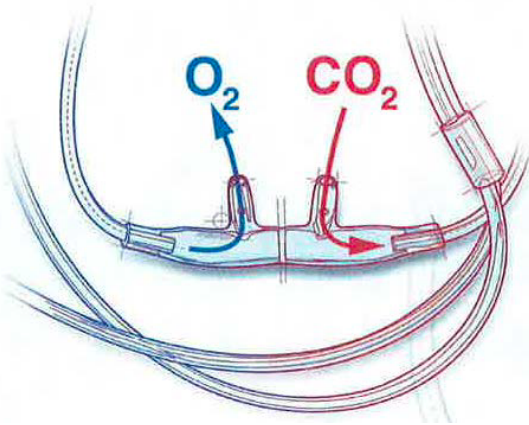


Figure 3: Drawing of the EtCO₂ divided sampling cannula with simultaneous oxygen delivery (Salter Labs, Arvin, CA). Instead of sampling CO₂ from the right nasal prong, we will measure nasal pressure.

cannula. The lower plot shows the airway flow measured during inspiration and expiration by a research spirometer

(RSS100 Korr Medical Technologies, Salt Lake City, UT).

Research Design:

Our research plan is to first, build a working prototype. Second, measure the device's performance using a simulated patient in order to identify the optimum pressure thresholds for the detection of a dislocated nasal cannula and the enunciation of alarms for inadequate tidal volume and respiratory rate. Third, to conduct clinical trials to test the feasibility of using our device during operative procedures.

Methods:

Aim 1: Build a working prototype:

Figure 5 demonstrates a future iteration of the prototype. Our device will connect

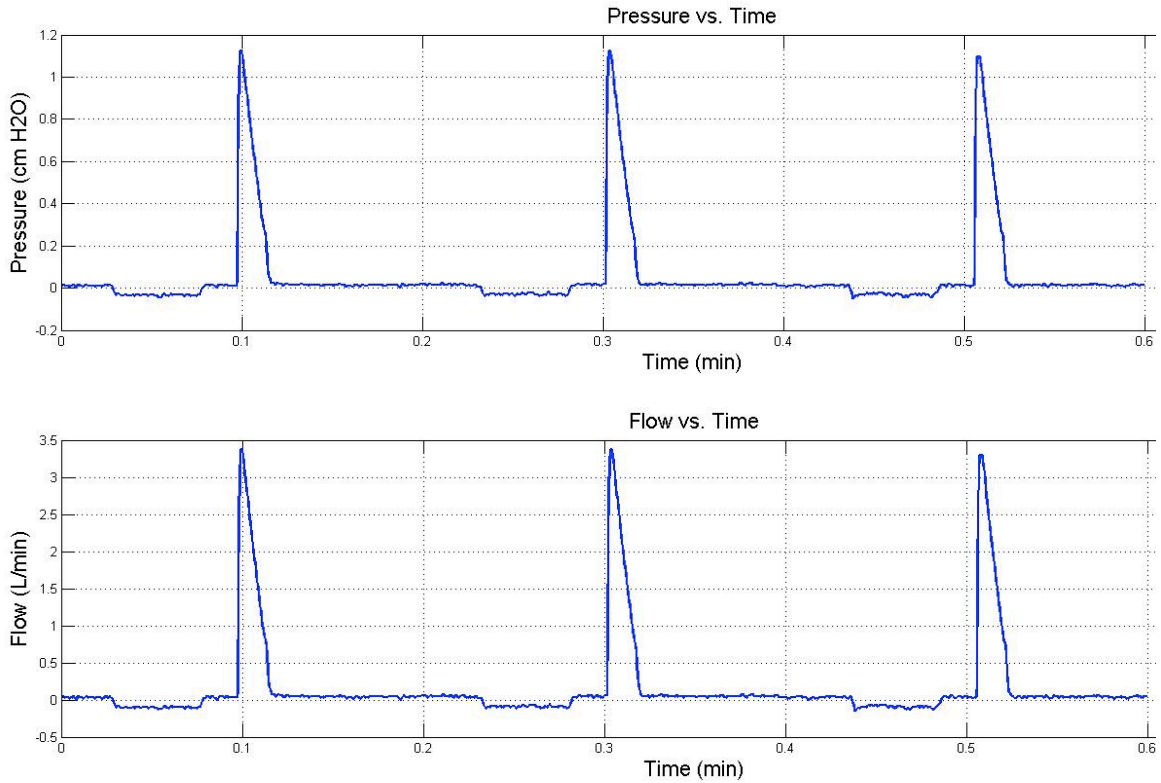


Figure 4: Top: Pressure measured in the right nasal prong of the cannula with 5 L/min of O₂ was delivered through the left prong. Bottom: Simulated airflow rate measured in the trachea during three spontaneous breaths.

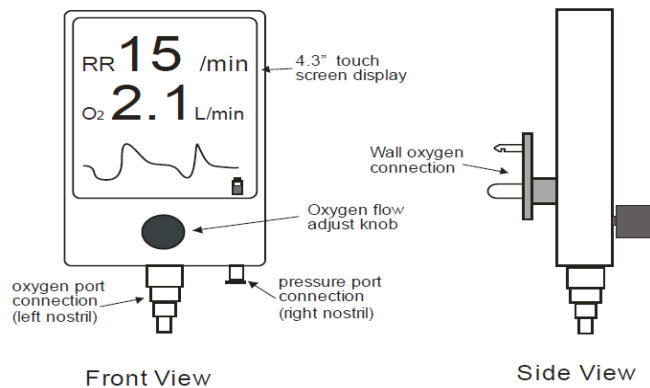


Figure 5: Future rendition of the prototype. It will include a touch screen display, connections for oxygen flow, pressure sensing and an oxygen flow control knob. The system will be powered by a rechargeable 7.2 volt 2200 mAh Li-ion battery. Computing will be provided by an AtMEGA 2560 microcontroller.

directly to a hospital oxygen connector at approximately 55 PSI. Due to the fact that built in wall pressure is much higher than the pressure at which supplemental oxygen is delivered to the patient, a precision flow resistor is necessary to decrease the pressure to a suitable 5 PSI. We will use a differential pressure transducer (MPXV5004DP, FreeScale Semiconductor, Austin TX) to measure the pressure drop as the oxygen flows through the resistor. A microcontroller will convert the pressure drop to a flow rate.

Software algorithms will be developed that use the oxygen flow rate to identify when a cannula is dislodged. Supplemental oxygen is delivered through the left nasal prong of the nasal cannula (Figure 3). As oxygen fills the patient's nasal cavity, the pressure within the cavity increases proportionally to the oxygen flow rate. Since we can accurately calculate the oxygen flow rate from the wall connector, we can calculate the anticipated nasal cavity backpressure when the nasal prongs are correctly positioned within the patient's nares.

Should either a single nasal prong or both nasal prongs become dislodged, the nasal backpressure will be abnormally low. During the simulation testing we will determine the baseline pressure threshold that accurately identifies a displaced cannula. Furthermore when a cannula is detected as dislodged, the device will sound an alarm and display a text message to alert the medical staff. This will provide adequate warning before incidences of apnea or other hypoventilation occurrences.

The pressure signal apnea monitor will measure the nasal pressure through the right prong of the nasal cannula (Figure 3). This is directly connected to the pressure port connector shown in Figure 5. The internal pressure transducer (BLVR-L01D, AllSensors, Morgan Hill, CA) includes a second compensating solid-state diaphragm that adjusts automatically for the effects of gravity and temperature. This helps to prevent positional changes and temperature drift. The pressure signal will be processed using a precision adjustable gain scale differential amplifier and a high-resolution analog to digital converter system. This is necessary due to the small measurements on the order of 0.001 cm H₂O.

Additional software algorithms will be developed to use the measured pressure signal to calculate the patient's expired tidal volume and respiratory rate. These algorithms will provide thresholds that were determined during the simulation testing adequate tidal volumes and respiratory rates. We expect that our device will have a detection rate of 97%, as reported by the literature [19]. The device's alarms will provide user feedback when the tidal volume is less than 150 ml or when the respiratory rate is below 4 b/min. These notification alarms will improve the patient

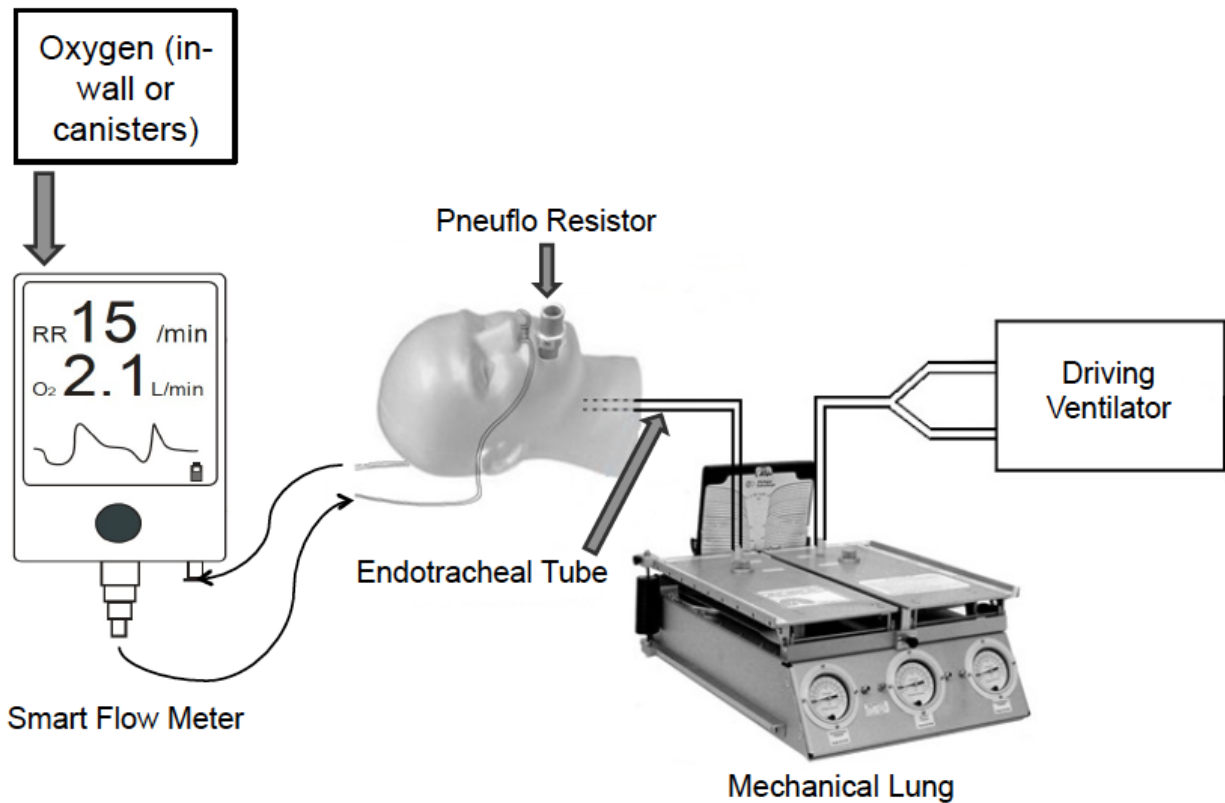


Figure 6: Visual representation of the system that will be used in both simulation testing and clinical trials. Includes a ventilator, two-chambered mechanical lung, mannequin, endotracheal tube, our pressure apnea monitor and supplemental oxygen. The flow meter will control delivery of oxygen to the patient. The pressure transducer measures nasal pressure through the nasal cannula. Pneuflo resistors vary the ratio of nasal to oral breathing.

outcomes and help prevent injuries caused by hypoventilation.

Aim 2: Simulation testing:

We will place a nasal cannula on a mannequin head as shown in Figure 6. Using anatomically correct nasal/oral cavities, designed to demonstrate varying sizes and shapes of human anatomy, we will simulate the air flow and pressure measurements. The 900C ventilator (Siemens Elema, Solna Sweden) will directly drive one chamber of the two chamber mechanical lung (Vent AID TTL; Michigan Instruments, Grand Rapids MI). A rigid metal strap will physically connect the two separate chambers of the mechanical

lung. When the ventilator delivers a tidal volume to the first chamber, the second chamber will rise, drawing in a tidal volume through the trachea of the mannequin, thus simulating a spontaneously breathing patient. The lung compliance will be set to 50 ml/cm H₂O and with an airway resistance of 8.2 cm H₂O/(L*s). The spontaneously driven lung will provide the respiratory measurements through a 7.0 mm endotracheal tube (Sheridan Oral *Nasal, Teleflex Medical, PA) to the mannequin. The ventilator will provide tidal volumes of 50, 100, 150, 300 and 500 mL and a respiratory rate of 1, 2, 4, 8, and 12 b/min. The accuracy by which the prototype delivers oxygen will be measured using a

gas flow analyzer (VT Plus, Fluke Biomedical, Everett, WA). Furthermore, we will use one of three Pneuflo resistors (Michigan Instruments, Grand Rapids, MI) in the mouth of the mannequin to partially occlude the oral/nasal breathing to simulate one of three levels (3.3 mm, 5.6 mm, and 7.7 mm). Additionally, we will test total airway obstruction by closing the oral section completely. During these simulations we will change the position of the cannula, moving the nasal prongs in and out of the nares. Also we will test the outcomes when one nasal prong is in the nose and the other is outside of the nose.

All outcomes will be analyzed and the device's performance will be determined by classifying each breath as one of the following: True Positive (TP), if a breath delivered by the ventilator is detected correctly by the prototype; False Negative (FN), if a true breath is not detected; False Positive (FP), if a breath is detected when the ventilator does not deliver a breath; True Negative (TN), if no breath is delivered and the prototype detects no breath. Sensitivity ($TP/(TP+FN)$) and specificity ($TN/(TN+FP)$) will be calculated as the detection threshold for each alarm is adjusted. This is done to find the optimum thresholds to provide accurate and reliable sensitivity and specificity. It is anticipated that our device will have sensitivity of 98% and specificity of 96%.

Aim 3: Human trial in volunteers:

In order to measure the apnea monitor's performance in a clinical setting we will recruit 40 volunteers from patients receiving procedures in the Gastrointestinal (GI) Lab of the University of Utah Hospital. Each patient will give prior consent to receiving a colonoscopy procedure using the approved

form provided by the University of Utah's Institutional Review Board. We chose to use the GI lab because Qadeer et al. reported that hypoxia occurs in 44-70% of all GI procedures [20]. With this being the case we anticipate to observe approximately 20 event of apnea.

In addition to our prototype, we will be using an end tidal CO₂ (EtCO₂) divided sampling cannula with simultaneous oxygen delivery (Salter Labs, Arvin, CA) that will be placed on the patient. During the procedure we will sample and record the SpO₂ (NM3, Philips Medical), tidal volume and respiratory rate via respiratory inductance plethysmography chest bands (Q-RIP, Braebon, Ogdensburg, NY) and via a disposable thermister system (Model 517, Braebon, Ogdensburg, NY). This reference data will be reviewed and analyzed; respiratory rates and tidal volumes will be calculated. Once the reference data is measured and annotated, detection of inadequate tidal volume and low respiratory rates, as measured by our prototype device, will be compared with the annotated reference data. Sensitivity and specificity will be calculated for the detection of cannula displacement, inadequate tidal volume and respiratory rates, as described in Aim 2. Once again it is anticipated that our pressure signal apnea monitor will have sensitivity of 98% and specificity of 96%.

Conclusion:

The use of a pressure signal apnea monitor has the potential to not only improve patient outcomes but to decrease overall health care costs. The device uses fundamental principles that have been previously proven accurate and reliable.

Future work will incorporate optimization of all software algorithms and hardware

performances and durability. Additionally, after all simulation testing and clinical data has been analyzed and reviewed the device will move from a laboratory prototype to a fully specified device for manufacturing.

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