SLEEP APNEA SIMULATION:
AN EXPERIMENTAL ASSESSMENT OF PHARYNGEAL COLLAPSE

May Cheng
Christopher Phaneuf
Brian Tovar
Department of Mechanical Engineering
The Cooper Union for the Advancement of Science and Art
New York, New York 10003

ABSTRACT

In this study, obstructive sleep apnea (OSA) was examined from the perspective of a physical airway model capable of simulating and measuring the behavior of the human pharynx. As a continuation of previous research based on a rigid model paired with computational fluid dynamics analysis consisting of flow through stationary airway boundaries, the goal of this experiment is to take a new step in the progression toward an instrumented, compliant airway model to parallel the development of a dynamic, virtual model. Our efforts yielded a collapsible airway model based on simplified geometry to capture the narrowed, highly flexible nature of the anatomy observed in conjunction with OSA. A data acquisition system was installed to interface with differential transducers for measuring pressures upstream and downstream of the airway and flowrate. Findings suggest a need for adjustments to the instrumentation and further model refinement before proceeding to the casting and testing of a realistic airway. The project has reached a milestone in the path toward reliable simulations and has established an intermediate step in the development of optimal methods for the realization of a meaningful simulation.

INTRODUCTION

Objective

The primary research objective is to advance the study of obstructive sleep apnea through the construction of an experimental flow apparatus capable of simulating the flow characteristics of a compliant pharynx subject to abnormal tissue geometry and variable collapsibility. The model should possess means of direct collapse observation, via materials with optical access and instrumentation for measurements at critical upstream and downstream locations. Material choices should comply with MRI compatibility criteria as well. A simplified airway will provide the foundation for future models possessing higher degrees of realism, detail, and variability. The physical model will serve to validate the results of finite element models developed in parallel to ensure proper correlations.

Background

An understanding of the basics of obstructive sleep apnea syndrome and the context of our research helps to highlight our motivations behind each design choice and allows a clearer interpretation of our results.

The Physiology of OSA. Obstructive Sleep Apnea (OSA) is a sleep disorder affecting flow through the human airway. Breathing is disrupted when adverse pressures cause the walls of the airway to collapse. In order for the classification of OSA to apply, as opposed to phenomena such as hypopnea, a measure of severity must be considered. Several indices are used to assess severity, including arousal or disturbance index, apnea-hypopnea index, and apnea index [1]. A simple definition of apnea is a temporary cessation of breath. More precise definitions prescribe conditions such as stoppage of breath or less than 25% of a normal breath for a period of 6 seconds or more [2]. A number of anatomic risk factors are associated with this disease and involve changes in the upper airway craniofacial and soft tissue structures, particularly the enlargement of the lateral pharyngeal walls, which studies have shown to be the most compliant tissue and a dominant risk factor among men [1,3]. The airway size...
is determined by the soft tissues of the pharynx (which include the tonsils, soft palate, uvula, tongue, and the lateral pharyngeal walls) and the craniofacial bone structures (which include the maxilla, mandible and hyoid bone). Factors that contribute to the changes in these structures are abnormalities in the upper airway structure, neck circumference, and obesity [1]. In addition to anatomy-based contributors, reduction in dilator muscle tone plays an important role in the collapsing behavior of the pharynx [3].

A separation of the pharynx into discrete sections can be summarized by the following division: 1) nasopharynx - nasal turbinates to the superior part of the soft palate, 2) oropharynx - soft palate to the epiglottis, and 3) hypopharynx - epiglottis to the larynx. The oropharynx is the most widely discussed segment, with statistics showing this area to be the site of collapse in 56-75% of OSA patients. Collapse at the base of the tongue is reported for 25-44% of patients and 0-33% experience occlusion of the hypopharynx. The shape of the pharyngeal cross-section is often an indicator of the likelihood of airway collapse. Normal subjects possess an elliptical airway with the major axis in the lateral dimension while snorers or sufferers of obstructive sleep apnea exhibit a more circular airway, sometimes forming an ellipse with the major axis in the posterior-anterior dimension [3].

Previous progress and related research. The ideas and goals of this project grew from past research by Dr. David Wootton at Drexel University. Students constructed an airway model to study the effects of airway geometry on internal airway pressure in children. A 3D model of the airway was generated from MRI scans. This model was meshed for use with computational fluid dynamics (CFD) simulations. Inspiratory pressure drop was found to be strongly correlated to area restriction and proportional to the square of the flowrate. A rigid airway based on the MRI-derived virtual model was manufactured using stereolithography rapid prototyping to validate the CFD model results. A test bed was assembled to conduct experiments to simulate flow through the rigid model. One of the critical components used to drive flow was a computer-controlled pump [5].

MATERIALS AND METHODS

Breathing Branch

The breathing branch of the test bed is comprised of the computer, the motor/pump, and the airway and involves controlling the flow of air through a flow apparatus simulating breaths through a human airway. The components for this branch were the first items approached in the process of building and instrumenting an experimental platform.

Computer-controlled pump. A computer-controlled pump, pictured in Figure 2, functions as our respiratory drive system. This critical piece of equipment was constructed during a previous phase of this study by students at Drexel. The primary components of this pump include a high-resolution stepper motor, motor controller, and a piston-cylinder assembly. Every part is rigidly mounted to an aluminum stand. The motion of the piston is operated through a motor controller that is programmed using a personal computer. IMS Terminal, a free program used for communication with the MicroLynx stepper motor module, was used to send instructions to the controller and execute a series of tasks. The simulation of a typical breathing cycle and the ability to vary flow parameters was developed through extensive use of documentation and a great deal of trial and error.

Acceleration current and run current are set as percentages of the maximum value. After this acceleration period, the motor slews at constant velocity then decelerates until it reaches its desired position, or exceeds the limit of the leadscrew.

Collapsible pharynx. Paramount to the success of this endeavor is the production of a compliant segment analogous to the collapsible human airway of an OSA sufferer. Fabricat-
ing this defining component is a formidable task for any engineer. The many unknowns and design considerations faced in approaching this key point led to the concept of an intermediate model to test with the entire system prior to final stages of research involving the detailed airway model. This intermediate model represents a simplified interpretation of the narrowing geometry of the oropharynx that contributes to pharyngeal collapse. By addressing the uncertainties of the casting processes without potentially sacrificing the time and material required by a final model, the difficult-to-anticipate issues are solved and the later phases of the project are able to run more smoothly.

Modeling a part of the human body and simulating the complexities of its operation demands the use of materials that will last under the stresses of repeated tests and behave true to their anatomical analogs. To match the behavior, the mechanical properties of the upper respiratory tract were taken into consideration. Reflecting characteristics of the human pharynx, the model is designed to handle pressure between -20 and 20 cm H\(_2\)O and exhibit a modulus between 5 and 25 kPa. The tissue substitute chosen was Sylgard 527, which consists of polydimethylsiloxane (PDMS), a dielectric silicone based organic polymer with stiffness in the relevant range of biological mechanical properties. A 1:1 component A:B mix of PDMS loaded at 0.0006s\(^{-1}\) resulted in a Young’s modulus of about 6 kPa [6]. Some of its advantageous characteristics include ease of preparation, transparency, low electrical conductivity, low polarity, elasticity, and resistance to swell in contact with water [7].

The mold for the preliminary airway was designed with 3-D mechanical design software SolidWorks. It consists of two basic components: an outer shell in the form of a tube and a two-part mandrel. The mandrel is a simple cylindrical shape that narrows in the middle to form an ellipse to create the hollow core of the resulting casting. One part of the mandrel features a base and a small square peg. The other half features the square hole at the narrow end to fix the pieces together and prevent rotation. Also included in the design is a round peg at the exposed end to allow easy removal once the material has cured around the mandrel. The mold produces a casting with an initial inner diameter of 2 cm, tapered to the ellipse with major and minor axes measuring 1 cm and 0.5 cm, respectively. The entire airway measures a little over 5 inches long and provides walls less than a centimeter thick. The two-part mandrel was made with a rapid prototyping machine that uses fused deposition modeling (FDM) to create plastic parts.

Incorporating the airway segment into a flow apparatus demanded a fitting and piping to connect the series of parts in line with the pump. Concerted effort was made in designing the airway model to minimize the number of part junctures, in turn reducing the likelihood of a leak. It was decided that mounting the compliant pharyngeal segment would require barbed fittings to be pressed into each end and secured to the connecting pipe. Early efforts to acquire fittings looked to distributors such as Cole-Parmer for a fitting that would meet the geometric specifications of our model. No commercial fittings met our needs and we decided to fabricate custom fittings, turned down on a lathe from a cylindrical stock of Delrin plastic. Machining the fittings allowed us to join them to the piping with press fits, side-stepping the need for threads and allowing better control over the area change at the transition between pipe and fitting. Hose clamps were secured over the ends of the airway to help ensure a seal.

It was initially planned to use a T-fitting for pressure measurement but the undesirable effects of the change in area at the fitting led to an alternative solution. A hypodermic needle would serve as the pressure tap, requiring a minute hole just smaller than the outer diameter of the needle. This was managed using a vertical milling machine at its highest spindle speed and a 1/64th-inch drill bit. Unfortunately, the first taps drilled into the PP pipe, which is used to connect every element in the apparatus, were not usable. Despite a tight fit with the needle, a burr had developed on the interior of the pipe due to its soft plastic properties. To alleviate this problem, the taps were instead drilled into the Delrin fittings. This harder plastic did not experience burr formation and ensured precise pressure readings. The PP pipe was then cut down to remove the portion with the burred hole.

**Nasal resistor.** The first airway structures encountered on inspiration are the nasal passages. Our model takes this critical feature into account by including an interchangeable orifice in the flow path. The simplification of this relatively complicated section of airway anatomy is justified by the notion that nasal passage collapse is not an issue when considering the pathogenesis of obstructive sleep apnea. Without any need to model dynamic behavior, only the effective pressure loss is reproduced. The nasal resistor component is comprised of a modified plastic coupling and a set of interchangeable, aluminum discs, each
characterized by the theoretical pressure drop corresponding to the diameter of the hole in its center. Three discs were fabricated using a standard lathe to drill three differently sized holes with diameters of 3, 5, and 6 mm. These values were calculated using a simple, theory-based equation dependent on pipe diameter, desired pressure drop, and flowrate (Eq 1). The availability of drill bit sizes limited the resolution of pressure drops achieved with the orifices. A three-part coupling for like-sized tubes facilitates the ability to swap the orifices. A vertical milling machine was used to modify the coupling’s o-ring seat to accept the custom nasal restrictor disc. This entire assembly is fitted directly on top of the pneumotachometer.

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\Delta p = \frac{8\rho Q^2}{\pi^2} \left( \frac{1}{D_2^4} - \frac{1}{D_1^4} \right)
\]

(1)

**Measurement Branch**

With the elements of the breathing branch in place and pressure taps ready for use, the measurement branch stems from three key positions on the airway model to a set of sensors running to a data acquisition system built into a personal computer.

**Sensors.** Three sensors were integrated into the setup: two for measuring static gage pressure at the upstream and downstream points along the airway and a third to measure flowrate. Despite the diversity of the measurement types, all three sensors are differential pressure transducers. Each is designed to operate over a different range, depending on its use. Since we are focusing on inspiratory flow, in which upstream conditions do not experience as extreme of a range of pressures, the transducer positioned upstream measures a range of ±2.5 in H₂O (±6.35 cm H₂O), while the downstream transducer handles ±10 in H₂O (±25.4 cm H₂O). The third transducer, made by Validyne, is specified for a smaller range of ±2 cm H₂O. The pair of Setra transducers are capacitance-based, powered by 9-30V and made to output 0-5V. The Validyne relies on magnetic reluctance, requiring a signal demodulator to output ±5V.

The two Setra transducers and accompanying power supply were mounted on a raised, angled platform for easy accessibility and stability. These transducers are each connected to one of the two pressure taps on the upstream and downstream sides of the airway. The static pressure tap consists of a hypodermic needle that was shortened using a belt sander. The procedure required angling the needle tip to avoid adverse bending and partial closure that results from shearing the thin-walled needle at a right angle. This eventually yielded a clean tip to be inserted into the drilled Delrin fittings, sealed with hot glue, and coupled to the transducer tubing with a barbed luer fitting. It is important to note that the interior finish of the drilled fittings was made as clean as possible and that the tips of the needles did not extend into the path of flow. The low-pressure Validyne transducer was used to record the pressure drop across a pneumotachometer, a device specifically designed to induce a resistance to flow that varies predictably with fluid velocity.

**Calibration.** Calibration of each pressure transducer was performed using a manometer. Pressure was applied to the transducer using a syringe. Using either an oscilloscope or the DAQ voltage measurement running continuously, adjustments were made to the zero and span settings. Analog voltage signals are read using a PCI-6120 DAQ Card channelled into an interactive user interface within National Instruments LabView. Data points were then correlated using a linear regression. In this instance, we used Microsoft Excel as our linear regression tool. The coefficients of the line of best fit were transferred to a formula module.
for collecting dynamic pressure data. For converting the pressures measured from the pneumotachometer into flowrates, calibration data supplied by the manufacturer produced an additional conversion formula. LabView reads the three sensor channels simultaneously and a formula function converts the measured voltage data to pressures and flowrates in real-time, logging them in a measurement file and plotting the results in a series of convenient charts. These charts can serve dually as a presentation medium and a debugging tool.

**Data collection.** To measure flow and pressures, we used National Instruments LabView Software interfaced with the PCI-6120 DAQ Card available from the same company. This particular card can convert four channels simultaneously with 16-bit resolution at one million samples per second. Our system design allows the user to set the number of samples and the sampling rate, either in the DAQ Assistant dialog box located in the block diagram or from the Data Collection tabs in the front panel. After the analog voltage has been converted into a digital signal at a certain sampling rate, the stream of numbers is logged into a data file using a **Write To File** module. The stream consists of 3 channels of data. By default, LabView logs a time variable nested within this stream. To prevent these unwanted data from being recorded, we have enabled the **Empty Time Column** setting from within the **Write To File** dialog box.

**RESULTS**

Measurements of flowrate and upstream and downstream pressures were recorded for trials of varying orifice sizes and varying pump speeds. Every data set was collected by starting the pump on an expiration stroke using IMS terminal, switching to LabView and initiating data collection at the best perceived moment at which the inspiration stroke began. The plotted data demonstrates that although a small segment of the first inspiration cycle was not measured (most likely due to software lag), the offset is repeatable and the waveform alignment is acceptable for plots of changing nasal resistance with pump speed held constant.

Data for the smallest (3 mm) orifice displays radically different behavior when compared to the larger orifices. The downstream measurement exceeded the lower limit of pressure measurement for the largest nasal resistance due to total collapse. Changing the rate of breathing has similar effects on the data, where slower pump speeds produce stable flow and higher speeds induce more extreme pressures and erratic flowrate characteristics.

Observation of the airway during pump operation provides visual confirmation of the behavior indicated by the data and helps interpret the physical significance. The fluctuating radial displacement of the airway walls is a function of flow direction, speed, and nasal resistance. Inspiration induces inward flexure of the airway, the magnitude of which increases with increasing speed and nasal resistance until reaching complete collapse at the smallest orifice (Fig 5). The airway flexes outward on expiration, swelling (or ballooning) more as pump speed and nasal resistance increase.

**DISCUSSION**

The plots of increasing nasal resistance (i.e. smaller orifice diameter) illustrate the cascading effect of pressure loss through the nasal passages. Higher nasal resistance causes lower pressures further downstream until a small enough restriction upstream causes the narrowing downstream to produce pressures at or below $P_{crit}$. This critical pressure is higher for sufferers of sleep apnea; it is often measured as a small negative or even positive pressure depending on severity while a normal sleeping subject will display a large negative critical pressure. This is what necessitates the use of continuous positive airway pressure (CPAP) masks, supplying the airway with enough pressure to compensate for the losses typically experienced while sleeping. Past experiments have used sleeping subjects suffering from OSA to study critical pressure by slowly dialing down the amount of pressure delivered by a CPAP mask until the subject experiences an apnea.

Flow limitation is an important phenomenon to observe as the indicator of partial or complete collapse. The data provide a clear look at the dynamics of choked flow. The transmural pressure, represented by downstream pressure, reaches a critical magnitude in which the airway closes and the flowrate fails to reach the same maximum seen on expiration. It is clear upon inspection that the results exhibit non-linear behavior (Fig 6(b)), resembling a hysteresis curve. The path leading up to inspiration does not identically match the path returning from inspiration. This is the case for the trial where collapse occurs, and for the trial
where it does not. These results are expected given the compliant characteristics of airway.

Many limitations hinder the current experimental setup from producing especially meaningful and publishable results. The intermediate model provides a general idea of the flow characteristics observed in the human airway; however, this simplified model is primarily meant to serve the intermediate purposes of debugging and refinement of the experimental platform.

The three differential pressure transducers proved adequate for low pump speeds (i.e. low flowrates) but fell short of capturing the full range of pressures corresponding to an average adult experiencing sleep apnea. The first issue that came to our attention was the limit of the Validyne transducer in measuring flowrate through the pneumotachometer. As previously explained, this device is designed to consistently induce a change in pressure that varies predictably with fluid velocity. When it was determined that the Validyne transducer would be insufficient for the measurement of airway pressures, it was relegated to dedicated use with the pneumotachometer, for which it was assumed to be perfectly suited. It was later realized that the maximum measurable pressure of 2 cm H₂O corresponds to peak flowrates of a little over 300 ml/s. Our findings indicate that this is useful for representing children and younger individuals but falls considerably short of allowing the simulation of adult breathing rates during sleep, which are measured with average peak flowrates around 400 ml/s.

The orifice sizes used to perform a comparison of a range of nasal resistances was limited by the narrow selection that we machined. The diameters are based on hand calculations from a theoretical formula (Eq 2) and correlate roughly with the observed resistances obtained from isolated experiments involving flow through only the resistor and past a single pressure tap. The original plan was to fabricate a minimum of five different orifices but drill bit availability reduced the options for diameters. A better approach would be an empirical assessment of a large series of orifice sizes to ensure knowledge of the anticipated pressure drop before the experiment.

REFERENCES