

A Passive Quantitative Measurement of Airway Resistance using Depth Data

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Abstract—The Respiratory Syncytial Virus (RSV) is the most common cause of serious lower respiratory tract infections in infants and young children. RSV often causes increased airway resistance, clinically detected as wheezing by chest auscultation. In this disease, expiratory flows are significantly reduced due to the high resistance in patient’s airway passages. A quantitative method for measuring resistance can have a great benefit to diagnosis and management of children with RSV infections as well as with other lung diseases. Airway resistance is defined as the lung pressure divided by the airflow. In this paper, we propose a method to quantify resistance through a simple, non-contact measurement of chest volume that can act as a surrogate measure of the lung pressure and volumetric airflow. We used depth data collected by a Microsoft Kinect camera for the measurement of the lung volume over time. In our experimentation, breathing through a number of plastic straws induced different airway resistances. For a standard spirometry test, our volume/flow estimation using Kinect showed strong correlation with the flow data collected by a commercially-available spirometer (five subjects, each performing 20 breathing trials, correlation coefficient = 0.88, with 95% confidence interval). As the number of straws decreased, emulating a higher airway obstruction, our algorithm was sufficient to distinguish between several levels of airway resistance.

I. INTRODUCTION

A. Motivation

Obstructive pulmonary diseases are characterized by airflow limitations due to the abnormal inflammation in the patient’s airway passages and/or constriction of bronchi or bronchioles that lead to an increase in the airway resistance and greater breathing difficulties. Two such diseases, affecting both children and adults, are asthma and the Respiratory Syncytial Virus (RSV). Asthma is a chronic condition while RSV is a major cause of lower respiratory tract infections in young children and infants. With older children and adults, the effects of RSV are milder [1]. In the United State, nearly all of the children are going to be infected with RSV at least once, by the end of two years of age [2]. Of those infected, about 2-3% will develop bronchiolitis and need to be hospitalized [3]. Children who have not been treated in

time, and developed bronchiolitis from RSV are at a higher risk of developing asthma later.

Many hospitals and clinics can rapidly test for RSV using a sample of fluid taken from the nose with a cotton swab [4]. Some cold-like symptoms (e.g. cough, stuffy nose, or low-grade fever) can easily be mistaken for RSV infection and cause a lot of unnecessary visits to hospitals and emergency rooms. RSV spreads easily by direct contact, so, these visits allow for rapid child-to-child transmission to previously uninfected children. In addition, this test only evaluates the presence of the virus and not the severity of the symptoms or the effectiveness of the current treatment regimen.

In this paper, we introduce a cost-effective and easy to operate system for passively measuring airway resistance during spontaneous breathing. Our approach uses signal processing techniques to infer airway resistance relative to a per-subject baseline using a commercially available infrared depth-sensor, the Microsoft Kinect. At a similar price to baby monitors, this method has the potential to greatly improve the treatment and diagnosis of infant pulmonary obstructive diseases and reduce unnecessary hospital visits. At present, we have only tested our method on adult subjects, but we expect to extend this approach to infants in the near future.

B. Related Works: Non-Invasive Respiration Monitoring

The diagnosis of the obstructive pulmonary diseases is usually confirmed by spirometry and measurement of the Forced Exhalation Volume (FEV1). Spirometry test is the measurement of the flow and volume of air entering and leaving the lungs [5]. In this test, the patient is asked to hold his/her breath and then exhale through the spirometer as hard as possible and the exhaled volume after 1 second is measured as the FEV1 parameter. Average values for FEV1 in healthy people depend mainly on sex and age, and a lower than the normal values confirms the presence of the airflow limitation and consequently a high airway resistance. Since infants and young children are unable to follow the instruction for the spirometry test, it is inappropriate for them. On the other hand, whole-body plethysmography is a technique for measuring lung volume and pressure passively in the research settings, however, its use for infants has been limited for research purposes and not for clinical management of the patients [6].

The abundance of low cost consumer electronics that can track/monitor the user without the need to attach a large number of sensors or markers to the body, opened up new opportunities for pervasive monitoring of pulmonary function in home setting. A promising approach to estimate the lung

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capacity is proposed in [7]. Authors used a microphone and a smartphone to capture and analyze acoustic signal during inhalation and exhalation to assist lung cancer patients regulate their breath. Despite the ease of use and the low price, the results can be effected by environmental audio noise.

In the last couple of years, several groups started using the Microsoft Kinect technology for non-contact respiratory motion measurement, which was released in November 2010, as an affordable motion-sensing device that can also capture depth data [8]. 3-D depth data is collected using infrared light, leading to the major advantage of the sensor being operational even in complete darkness almost independent of the lighting condition in order to monitor patients/infants during their sleep. Authors in [9], investigated the Kinect measurement performance based on the distances from camera for potential motion tracking applications. They used a spline surface model to track the chest motion during respiration [10]. A real-time respiratory monitoring system using Kinect was also developed in [11], that tried to enhance the inherent depth resolution of the depth camera via a simple motion magnification approach. Meanwhile, various methods of respiratory volume calculation using the Kinect depth data have been suggested [12], [13], [14].

These studies attempt to measure respiration rate, air-flow and lung volume non-invasively and without having the patient follow any commands. Low peak airflow is consistent with the obstructive pulmonary diseases, but has poor specificity because it can be caused by other lung diseases and can be altered by the subject unintentional interference. Therefore, measuring airflow by itself is not a reliable estimation of the airway resistance.

C. Our Contribution

We present a novel use of the Microsoft Kinect camera to create the first inexpensive quantitative measurement of airway resistance for patients that cannot administer to the spirometry particularly the infants. The system accurately captures the breathing patterns, airflow and lung volume using the Kinect data. By employing a mathematical model of lung mechanics, airway resistance is then estimated from a Least Squares (LS) solution of a well-known airway resistance equation [15]. For the experimentation, different airway resistances have been induced by breathing through a number of straws. Our system measures airway resistance of one breathing scenario relative to another for the same subject and ultimately can be used to demonstrate the relative improvement in airway obstruction before and after pharmacologic treatment.

II. METHODS: PASSIVE AIRWAY RESISTANCE MEASUREMENT

Airway resistance is the mechanical cause of most symptoms in obstructive pulmonary disease and can be considered the primary measure of disease severity. Airway resistance, R , is defined as:

$$R = \frac{\Delta P}{Q} = \frac{P_{air} - P_{lung}}{Q} \quad (1)$$

where Q is the airflow, P_{lung} is the pressure inside the lung (intrathoracic) and P_{air} is the atmospheric pressure that can be considered constant for a given altitude [15].

Directly measuring P_{lung} requires a pressure sensor placed deep in the lung, which is generally impractical. We will use a biomechanical model to infer lung pressure from lung volume, as explained in Section II-B. The airflow, Q , is easier to measure. We use depth data from Kinect to estimate relative lung volume. The derivative of this data is the airflow.

A. Flow Estimation Using Kinect

If accurately captured, the expansions and contractions of the patient's torso can be used to estimate the lung vital capacity and the respiratory rate. The resolution of the Kinect depth data allows for measurement of changes in depth down to millimeter accuracy [9]. Kinect can accurately capture the rise and fall of the chest region during each breath cycle. The lung volume, $V_{lung}(t)$, at time t can be estimated by numerically integrating the depth value, $d_{ij}(t)$, of every pixel in the region of interest, \mathcal{R}_{chest} .

$$V_{lung}(t) = V_0 - \sum_{i,j \in \mathcal{R}_{chest}} d_{ij}(t) \quad (2)$$

where V_0 is the volume between the subject's chest and the camera. In our study, the chest is a rectangular region bounded by the "right shoulder", "left shoulder", "right hip", and "left hip" as shown in Fig. 1a.

Airway flow rate, $Q(t)$ can be calculated as the volume change in the chest or more exactly, the thoracoabdominal region per time unit. This is the derivative of $V_{lung}(t)$. Using numerical techniques, the flow is calculated as follows;

$$Q(t) = \frac{dV_{lung}}{dt} \approx \frac{\Delta V_{lung}}{\Delta t} = \frac{V_{lung}(t) - V_{lung}(t - \Delta t)}{\Delta t} \quad (3)$$

where $Q(t) > 0$ during inspiration and $Q(t) < 0$ during expiration. Since numerical differentiation is susceptible to noise, it is often necessary to filter the data or curve fit it to obtain better results. The $V_{lung}(t)$ used in (3) is smoothed using a moving average filter of size 5.

If the subject distance from the camera changes between trials, then the value for V_0 would not be consistent. Further, for each experiment, a different bounding box is used for \mathcal{R}_{chest} . To correct for this and improve consistency between subjects, we normalize the $V_{lung}(t)$ data so all values lie between 0 and 1: a value of 0 refers to a deflated lung, and a value of 1 is a fully inflated lung. Fig. 1b shows the volume waveform extracted from Kinect depth data using (2) along with the filtered signal and the airflow extracted using (3) during three normal breathing cycles.

B. Lung Pressure Estimation

Pressure in the lungs, P_{lung} is related to the elastic recoil of the lungs, which is also correlated to the volume of the lung by its volumetric Young's modulus. This relationship

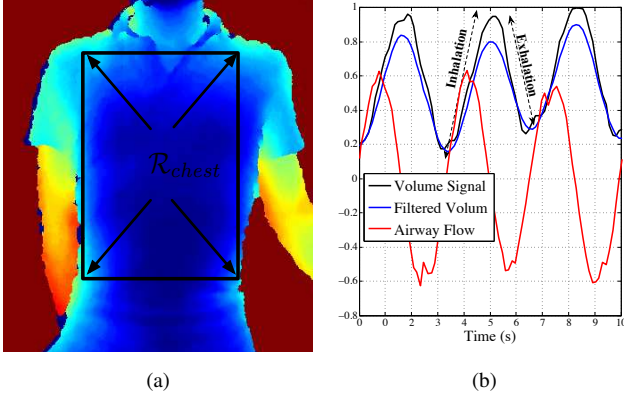


Fig. 1. (a) Chest bounding box on a color coded depth image. (b) $V_{lung}(t)$, filtered $V_{lung}(t)$ and $Q(t)$ extracted using Kinect depth data during normal breathing.

has also been verified by curve fitting of the results generated for the human lungs by simultaneously measuring changes in the lung volume with a spirometer and changes in pleural pressure with a pressure gauge [16]. Given a baseline resting volume at the end of an expiration, and considering a piecewise linear lung pressure-volume relationship [17], the negative intrathoracic pressure can be estimated from chest geometry:

$$P_{lung} \propto V_{lung} = k_1 \cdot V_{lung} \quad (4)$$

Maximum pressure is developed at maximum lung volume. Note that the pressure-volume model for lung has different parameters for inspiration and expiration periods. The volume in the lungs, V_{lung} can be calculated from measuring the geometry of the chest (e.g. chest circumference considering a cylinder model for thorax) or using (2). This model was derived for spontaneous (normal) breathing, although we will show experimentally that it fits quite well for forced breathing (FEV1) as well, albeit with a different k_1 .

C. Least Squares Estimation of Airway Resistance

By taking the airway resistance equation, (1), and combining it with (3) and (4), we derive the following equation:

$$R = \frac{P_{air} - k_1 \cdot V_{lung}(t)}{\frac{\Delta V_{lung}}{\Delta t}} \quad (5)$$

By using this mechanical model of the lung, the calculation of R only requires the measurement of lung volume over time. Therefore, a passive non-invasive measurement of the lung volume while breathing will yield the airway resistance, R . A linear equation system can be built based on (5) over a breathing time interval, $t \in [0, T]$. A least squares approach is applied to calculate the unknown constants in the following equations:

$$\begin{bmatrix} \frac{\Delta V_{lung}}{\Delta t} & V_{lung}(t) \end{bmatrix} \begin{bmatrix} R \\ k_1 \end{bmatrix} = P_{air} \cdot I \quad (6)$$

where I is an identity matrix with the size of data samples in $t \in [0, T]$ interval. $V_{lung}(t)$ and $\frac{\Delta V_{lung}}{\Delta t}$ are calculated using (2) and (3), respectively and $[R \ k_1]^T$ is the vector of unknown parameters. Note that since the pressure-volume model for lung varies between inspiration and expiration, (6) should be solved for inspiration and expiration periods, separately.

k_1 that is the linear correlation ratio between the lung internal pressure and the lung volume, remains relatively constant for each individual but with different values for inspiration and expiration. Therefore, (6) can be solved for a sample breathing trial and the estimated k_1 s for inspiration and expiration then fixed as a subject-specific parameter for the future experiments.

III. EXPERIMENTAL RESULT

A. System Configuration

We used the Microsoft Kinect [8] for 3-D depth data and SP10 spirometer (ContecTM, Qinhuangdao/China) [18] for respiration data recording. Kinect depth data was recorded at 30 frame/sec with the resolution of 640×480 pixels per frame. The SP10 spirometer has the flow range between $1 L/s$ and $16 L/s$ with the maximum volume measurement of $10 L$.

B. Experimental Procedure

In this study, a series of experiments was conducted at the GT-Bionics lab, Georgia Institute of Technology. Five healthy subjects including one female with ages from 19 to 30 year old participated in this pilot study. All participants signed an informed consent form prior the experiment.

Our experiment had two sessions: forced breathing session and spontaneous breathing session. For both sessions, the Kinect was placed on a table in front of the subject such that the center of the depth sensor was placed at a height of $1.275 m$ from the ground and $1.14 m$ away from the wall where the subjects leaned against during the measurements.

For the forced breathing session, initially the subject was asked to practice performing the spirometry test using a commercially available spirometer, SP10 [18]. The spirometer measures the FEV1 as well as Forced Volume Capacity (FVC), which is the entire volume that can be breathed out during maximal expiration. In this session, only the exhaled volumes were recorded with the spirometer as well as the chest movements with the Kinect. To emulate the effect of airway obstruction on spirometry test, three mouthpiece adaptors were made for the spirometer, with one, three or five straws (each with diameter of $4.5 mm$). Each spirometry test with and without mouthpiece adaptors was repeated five times.

For spontaneous breathing session, the subject pressed his/her back against the wall and breathed normally while chest movement was recorded by the Kinect depth camera. Furthermore, the subject moved a wireless mouse with the right hand on the wall from left to right to mark the timestamps for inhalation and exhalation. This breathing experiment was repeated five times for normal breathing and

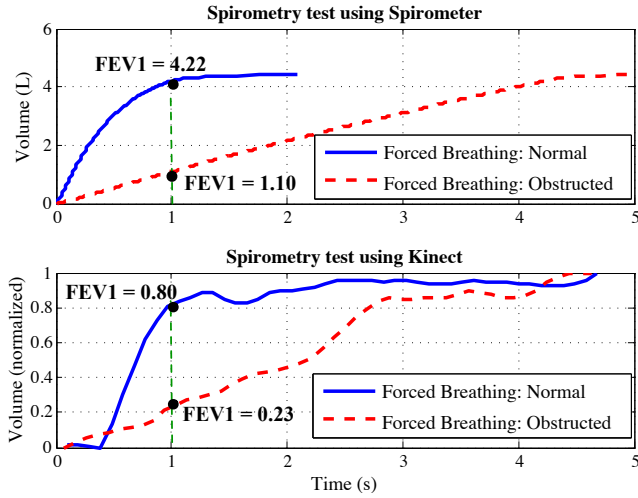


Fig. 2. Spirometry test result from SP10 spirometer (top) and Kinect (bottom) for normal and obstructed (using one straw) breathings. FEV1 parameter (volume after 1 second) for each test is shown.

then obstructed breathing through one, three, and five straws, each for one minute.

C. Forced Breathing Session

The output of spirometer and Kinect volume calculation for Subject 5 (male) for two different breathing experiments (normal and obstructed) is shown in Fig. 2. Both spirometer and Kinect data show a significant drop in FEV1 parameter, while the absolute value of the FVC stays almost the same in either experiment. This is because the subject already inhaled the air as much as the lung capacity allowed and by the end of the forced exhalation uses all of that capacity no matter how much resistance was in the airways. In the case of obstructed pulmonary diseases, FVC may also be reduced because gas is trapped behind obstructed bronchi but this reduction to a lesser extent than FEV1.

The average over five trials of FEV1 and FVC parameters acquired by spirometry test for four breathing conditions is shown in Fig. 3. The absolute values of FEV1 and FVC parameters were significantly lower for one of our subjects (subject 2). This subject is our only female subject in this experiment and the result confirms the gender dependency of human's vital capacity. For all subjects, as expected, FEV1 parameter is dropping as the number of straws decreases. This shows the higher the airway resistance (fewer straws), the slower the forced expiration airflow. However, this is not the case for normal breathing in comparison with obstructed breathing using straws for most of the subjects. One reason could be based on the psychological effect of using straws that makes the subject to apply more pressure and effort for forced exhalation.

To validate the correlation between volume detection using spirometer and Kinect depth data, for each subject, the correlation coefficient (CC) between FEV1 parameters recorded by spirometer and Kinect, along with the lower (CC_L) and upper (CC_U) bounds for a 95% confidence interval (CI) among five trails is given in Table I. Except for Subject

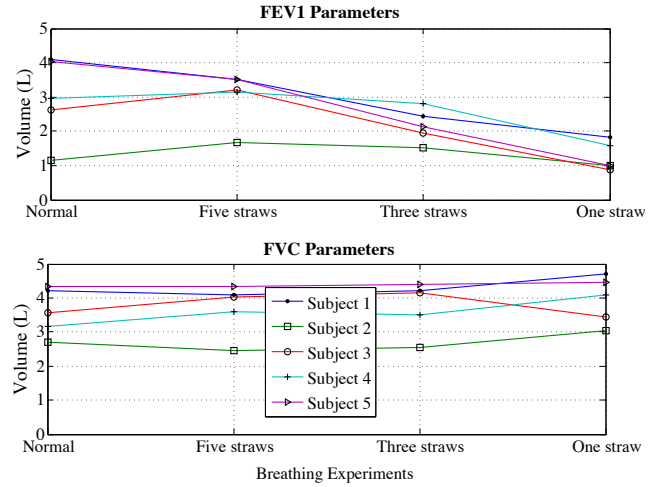


Fig. 3. Averaged FEV1 (top) and FVC (bottom) parameters for each subject from SP10 spirometer for normal and obstructed (using one, three and five straws) breathings.

2 (female), a very high correlation between spirometer and Kinect data is observed.

To estimate the airway resistance in spirometry test with different obstructions, in (6), we assumed $P_{air} = 1$ among all subjects and trials (average sea-level pressure is $P_{air} = 1 \text{ atm} = 760 \text{ mmHg}$). P_{air} is a constant value (no significant height change), and based on (6), its absolute value does not have any effect on relative values of extracted R . The solution of the overdetermined (6) is estimated using a least squares approach. First, we solved (6) for only one obstructed breathing scenario (breathing through three straws) and extracted the k_1 parameter during expiration. Then, using this fixed k_1 for a given subject (listed in Table II for forced session), we solved (6) to extract R for different obstructed breathing experiments. Fig. 4a shows the extracted R (averaged over 5 trials) of each subject for different breathing experiments in forced session. For obstructed breathing experiments, a semi-linear increasing pattern in R is observed as the number of straws decreases.

D. Spontaneous Breathing Session

In this session, subject is asked to breathe for a time interval of one minute that includes several breathing cycles based on the subject's rate of breathing. As shown in Fig. 1b, breathing cycles are accurately detectable using Kinect depth data. We also verified the start and end of each breath cycle using synchronized mouse data that confirmed the reliability of the depth data on respiration cycle detection.

TABLE I
CORRELATION COEFFICIENT (CC) WITH 95% CI BETWEEN FEV1
PARAMETERS EXTRACTED FROM SPIROMETER AND KINECT

Subjects	Sub 1	Sub 2	Sub 3	Sub 4	Sub 5
CC	0.95	0.75	0.90	0.86	0.96
CC_L/CC_U	.88/.98	.45/.89	.77/.96	.64/.94	.90/.98

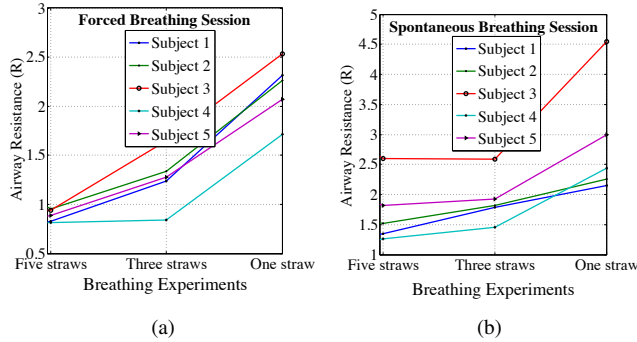


Fig. 4. Extracted airway resistance (R) for each subject for obstructed (using one, three and five straws) breathings during (a) Forced breathing session, (b) Spontaneous breathing session.

Here, like previous breathing session, we first extract k_1 parameter during expiration cycles for each subject, using data from obstructed breathing experiment with three straws. Table II shows the estimated value of k_1 for each subject in the spontaneous session. It was expected to have the same k_1 for a given subject independent of the type of experiments, since k_1 is a parameter correlating the volume and pressure of the subject's lung, however, Table II clearly demonstrates different values for the two experimental sessions. This outcome needs more investigation with more subjects in a series of controlled experiments.

Using k_1 computed for the spontaneous session, airway resistance R can be calculated based on (6). Fig. 4b illustrates the extracted R of each subject for different breathing experiments in the spontaneous session. As in Fig. 4a, this figure confirms that R increases as the number of straws decreases.

For three out of five subjects, the difference in value of R between breathing with five and three straws is much less than the difference between breathing with three and one straws. The fact that breathing through straws applies resistance in series to the normal airway resistance, can justify this outcome. For some subjects (here subject 3, 4, and 5), five or three straws are still in the range of their natural airway resistance, hence the increase in R was not noticeable and only appeared when using one straw which applied significantly higher resistance than their natural airway resistance.

TABLE II

AVERAGED k_1 PARAMETER DURING EXPIRATION FOR FORCED AND SPONTANEOUS BREATHING SESSIONS

Subjects	Sub 1	Sub 2	Sub 3	Sub 4	Sub 5
Forced k_1	0.85	0.91	0.87	0.81	0.88
Spontaneous k_1	0.94	0.95	0.73	1.03	0.88

IV. CONCLUSIONS AND FUTURE WORK

Periodic monitoring of airway resistance can provide a quantitative measure for emerging therapies against obstructive pulmonary diseases, such as RSV. In this paper, we proposed a new non-invasive and inexpensive approach to estimate airway resistance without active patient participation. The significant changes in resistance (induced by

breathing through a number of small straws) measured by our proposed method showed high correlation with data acquired from a spirometer. While, we had small samples from few subjects, the recorded data was used for model development and to obtain preliminary conclusions before study is expanded. A quantitative airway resistance measure, as an indicator of the disease severity, is valuable in studying vaccine effects with fewer patients, and may decrease the costs and risks of testing candidate vaccines. Furthermore, it would help clinicians to evaluate patients with acute illness, and determine those that are at risk of serious complications. This can reduce a lot of unnecessary visits to the hospitals and emergency rooms.

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