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Noninvasive Method and Metric for Monitoring Lung Condition*

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***Abstract*— For hospitalized patients with pulmonary conditions, the onset of respiratory decline can occur unnoticed, due to the absence of a way to continuously and noninvasively monitor lung condition. Based on the relationship between lung volume and pleural pressure, we hypothesized that the time delay (Δt) between the start of a respiratory cycle and the occurrence of lung sounds associated with inspiration would correlate with lung volume. Additionally, we developed a research tool, consisting of a respiration belt, digital stethoscope, data collection system and MATLAB algorithm, to measure this delay. We conducted a feasibility study with three healthy individuals that involved safely manipulating lung volume, through subject position and activity, and plotting Δt against volume measurements obtained via spirometry. The results indicated that Δt was measurable and changed with lung volume and, therefore, has the potential to serve as a lung condition monitoring tool.**

***Clinical Relevance*—Developing this metric and measurement method into a noninvasive monitoring tool would provide means to better detect declining pulmonary function and intervene earlier and more appropriately.**

I. INTRODUCTION

Auscultation is a standard tool for monitoring and assessing pulmonary condition. While it provides the experienced clinician with rich insights into respiratory mechanics, it is highly qualitative, dependent upon training and experience, and provides only discrete pictures of a patient's lung health [1], such that changes may not be detected in a timely manner. Patients receiving intensive or intermediate care would benefit from frequent monitoring to detect improvements or the onset of a reduction in lung capacity, enabling earlier, more effective interventions. Patients suffering from reversible changes in lung compliance, such as pulmonary edema [2], would also benefit from a reliable means to detect condition and assess the effectiveness of interventions, which can be as simple as ambulation or the administration of diuretics [2].

Spirometry, as an adjunct to auscultation, is also discrete, requires active patient participation and, unless filters are fitted, aerosolizes particles, which is a significant concern with infectious lung conditions. Some technologies, such as the StethoMe® smart stethoscope [3] and the Respira Labs' Sylvee [4], attempt to provide continuous and quantitative evaluation tools, however they lack the basis of using traditional passive auscultation to aid physicians in detecting respiratory deterioration. A key need is a single quantitative lung condition metric, equivalent to heart rate or blood oxygen saturation, which can be measured non-invasively and continuously, to provide information on trends in a patient's condition.

Our potential metric for lung condition was inspired by the clinical practice of observing the temporal relationship between the onset of inspiration and detection of respiratory sounds. Because early respiratory sounds are empirically linked to lung volumes—likely reflecting how the lungs store and dissipate energy during the respiratory cycle [5], [6]—these sounds are often referred to as "opening sounds." Clinical observations indicate that normal lung volumes have a short interval between the onset of inspiration and opening sounds, whereas underinflated lungs have a delay in opening sounds. Rather than interpret the sounds directly, we sought a link between the timing of these phasic sounds and lung volume. We hypothesized that the delay (Δt), on the order of tenths of a second, between the start of the diaphragm's contraction and a subsequent burst of sound (opening sounds), could be detected digitally and would correlate with lung volume, defined by tidal volume. Testing this hypothesis required developing an apparatus and algorithm to capture and evaluate Δt.

II. METHODS

A. Study Summary

This study's objective was to investigate the existence and measurability of Δt and its potential as an indicator of pulmonary function, as a proxy for tidal volume (TV). We identified and integrated off-the-shelf sensors to capture audio and the onset of respiratory cycles, over periods on the order of a minute. This would allow averaging to minimize the impact of transient fluctuations. In parallel, we developed a signal processing algorithm to measure Δt. With this hardware and software, we executed a protocol that used physical position and activity to safely change subjects' tidal volume, while measuring Δt and conducting spirometry to directly measure TV. Three team members served as subjects

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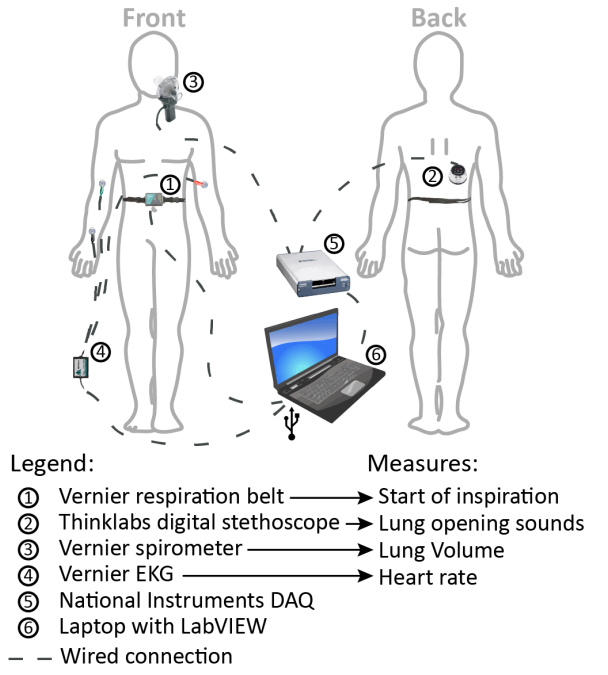

Fig. 1: Experimental Setup. The respiration belt and stethoscope are used to acquire the Δt. The spirometer and EKG sensor are used to verify subject condition. A LabVIEW program was developed to acquire the output signals from these sensors, verify synchronization, and write the data to a CSV file.

for the purposes of investigating feasibility, as a precursor to a formal study. (MIT IRB protocol 2107000447.)

B. Sensing Hardware

The sensors used in the experiments are described in the following and their placements are shown in Figure 1.

Respiration Belt: A Vernier Go Direct® Respiration Belt (Beaverton, OR) was chosen as a validated method to measure respiration. Other methods based on electromyography or accelerometer are noisy and sensitive to placement and will be considered in future work. The belt integrates a force sensor (120Ω strain gauge) and an adjustable nylon strap placed around the subject's diaphragm to measure respiration effort and respiration rate. The sensor has a force range from 0 to 50 N and a force resolution of 0.01 N. When connected via USB to a computer, it has a response time of 50 ms and a sampling rate of 20 Hz.

Digital Stethoscope: The Thinklabs One digital stethoscope (Centennial, CO) utilizes a patented capacitive sensing technology, which enables high-quality signal acquisition, amplifies the signal and optionally filters it with five preset filters. The 100 - 1000 Hz bandpass filter was selected, as it is effective for capturing normal lung sounds, while blocking lower frequencies and vibrations [7]. Further filtering was conducted by the algorithm. The stethoscope's 3.5 mm jack was used to connect it to the data acquisition (DAQ) system. The stethoscope was positioned in the ninth intercostal space, avoiding muscular tissue, to reliably capture lung opening sounds and held against the subject's back at a near constant pressure with an elastic garment.

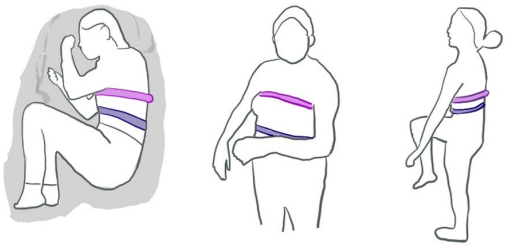

Fig. 2: Experimental positions including: fetal (left), standing (center), and running (right) positions. The stethoscope and respiration belt are located superiorly and inferiorly, respectively.

Digital Spirometer: Vernier's SPR-BTA Spirometer measures forced expiratory volume, forced vital capacity, and tidal volume. It was also used to validate the respiratory timing, identified in the respiration belt signal. The spirometer flow rate range is ± 10 L/s and has a nominal output of 128 mV/L/s. SparkFun's Vernier Interface Shield was used to connect the spirometer's analog voltage output to the DAQ.This was operated in handheld mode.

Electrocardiogram (EKG) Sensor: Vernier's EKG Sensor is a three-electrode system that measures cardiac electrical potential waveforms and determines heart rate by examining the number of QRS waveforms over a period of time. The sensor has a gain of 1mV body potential/1 V sensor output with an offset of approximately 1V ($\pm$.3V) [8]. This communicated to the PC via USB. The electrodes were affixed to the subject's arms as shown in Figure 1.

Data Collection and Synchronization: A National Instruments (NI) USB-6343 Multifunction I/O board was used to digitize the audio and spirometer signals. For this experiment, the sampling rate was set at 30 kSamples/sec and a voltage range of $\pm$10V. The NI board, respiration belt, and EKG communicated via USB with a PC running a LabVIEW. This set the acquisition rates, executing parallel data processing operations, logged each signal in a CSV format file, and verified the synchronization of the sensor data. A high-level diagram of the sensor connections is shown in Figure 1.

C. Experimental Procedure

The stethoscope, respiratory belt and EKG were fitted to each subject who was then asked to, in turn, assume a baseline standing position, assume a fetal position and then run in place, as shown in Figure 2. Each state was maintained for approximately 30 seconds, with the subject's heart rate monitored to verify that cardio-thoracic stability. (For the running condition a heart rate of $>$ 100 beats per minute was acceptable.) Once stable, the belt and stethoscope were sampled for 30 seconds and this measurement was repeated 4 times. Subsequently, the subject was instructed to breathe normally through a spirometer, while pinching their nasal airway closed, and their tidal volume recorded. We expected that comparing running to standing to the fetal position would evidence a natural decrease in lung volume and an increase in Δt.

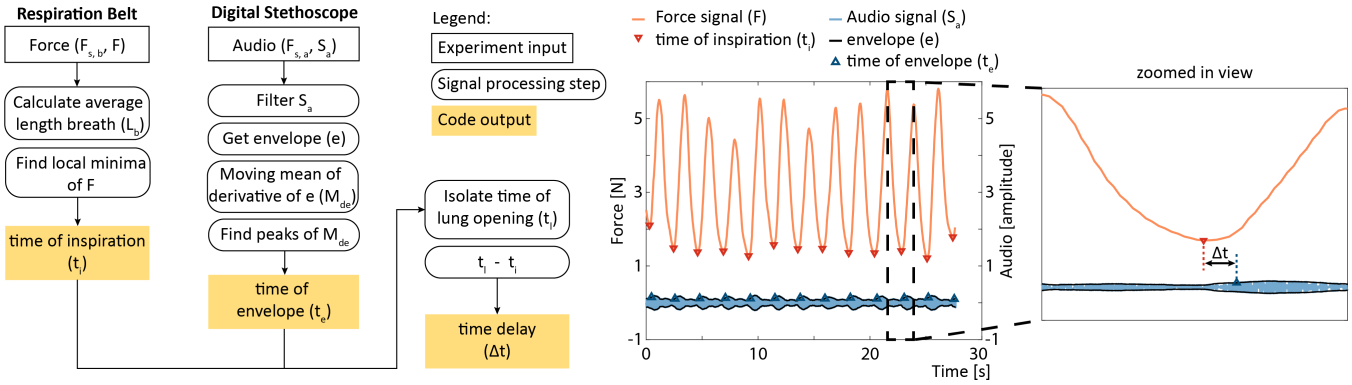

Fig. 3: Left: Block diagram of Signal processing used to determine the time delay (Δt) from time of inspiration (t_i) and the time of lung opening (t_l). Right: Force signal showing minima and audio signals showing points of sharpest increase, between which Δt is measured.

D. Signal Processing

The MATLAB script is represented with a block diagram in Figure 3 at left and sample signals shown at right. At a high level, for each breath cycle the code identifies the time of inspiration (t_i) from the force signal, envelopes the audio signal and finds the onset of lung opening sounds (t_e) and calculates the delay. These values are then averaged over the sample to report Δt.

The onset of respiration was defined as the time t_i at which the minimum force (F) was exerted on the belt. This is the moment just before respiration begins. To classify a single minima per breath cycle, a local minima finder was used with a window of $0.75\,F_{s,b}\,L_b$, where $F_{s,b}$ is the sampling frequency of the respiration belt and L_b is the average breath length. L_b was calculated from the average length of time between the peaks in F.

The onset of lung opening sound was defined as the time t_e of sharpest increase of the audio signal. First the audio signal S_a was further filtered using a 50 - 400 Hz bandpass filter, to isolate the lung sounds and remove large spikes of noise. Then envelope detection was then used on S_a to generate an envelope (e).The points of greatest slope were identified by taking a moving mean of the derivative of the envelope (M_{de}), using a window of $\frac{F_{s,a}}{2}$, where $F_{s,a}$ is the sampling frequency of the audio signal.The peaks of M_{de} corresponded to the points of sharpest increase or t_e. Multiple t_e were occasionally identified in a single breath cycle, for both inspiration and expiration sounds. The lung opening sounds (t_l) were isolated by selecting the t_e closest to the t_i.

The Δt is calculated as $t_l - t_i$. For the 30 second sample shown in Figure 3, approximately 13 Δt values were calculated, corresponding to the number of breaths taken by the subject. A sample with these Δt values, the average Δt, and the Δt ± 2 standard deviations can be seen in Figure 4.

III. RESULTS

The Δt values from each of the four 30-second tests were averaged and compared across the three subjects and three experimental conditions. Due to noise within the acoustic signal and variance within the time delay algorithm, the outliers in each 30 second sample were removed using the median method within the *rmoutliers* function in MATLAB. Figure 5 presents the average tidal volumes measured with the spirometer and the average Δt values for the standing, fetal, and running conditions for each subject.

A trend between the various subject conditions can be seen in Figure 5. The standing condition was selected as a baseline lung volume for the subject, and the fetal and running conditions correlate with a decreased and an increased lung volume state, respectively. The overall trend shows decreasing lung volume resulting in an increasing Δt. Although the Δt values between the baseline and decreased lung volume states vary minimally, there is a distinct deviation for the increased lung volume state from the baseline. Performing a paired t-test for the fetal and running conditions with the standing condition as the baseline corroborates the visual differences in Figure 5. Using a p-value of 0.05, the difference between the two conditions' means is statistically significant for the comparison between standing and running, as shown in Table I. The resulting p-values for the comparison between standing and fetal fall above .05, and thus are not statistically significant.

IV. DISCUSSION

The statistically significant decrease of Δt values from the baseline lung volume state to the increased lung volume state provides preliminary validation of the experimental hypothesis. The statistically insignificant comparison of Δt

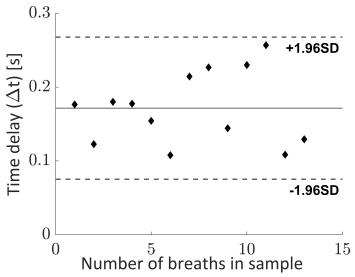

Fig. 4: The time delay, Δt (black diamond), for each breath in a 30 second experimental sample from the study. For each sample, the average Δt (solid line) and Δt ± 2 standard deviation (dashed lines) were found. The data points for this sample fall within 2 standard deviations of the mean, indicating a narrow distribution.

TABLE I: P-values from paired t-tests between Standing and Fetal/Running conditions for all subjects

Comparison with Standing	p-value
Subject 1 - Fetal	0.71
Subject 1 - Running	0.0022
Subject 2 - Fetal	0.39
Subject 2 - Running	0.045
Subject 3 - Fetal	0.11
Subject 3 - Running	6.2e-04

between the baseline and the decreased lung volume state is mainly attributed to the difficulty of reproducing a contracted lung volume within healthy subjects. Healthy lungs are more resilient to deflation than non-compliant diseased lungs, so the duration of the conducted fetal position tests may not have been adequate to mimic an unhealthy lung state and generate a change in Δt. A better way to achieve measurements at low TVs in healthy patients would be to gather data shortly after a subject wakes up from non-REM sleep, when volumes have been shown to be lower [9].

The variation of Δt within the baseline state was substantially greater than in the other two states, as shown by the error bars in Figure 5. We believe this is due to the more controlled nature of the fetal and running experimental tests, which forced the lungs to contract and expand, respectively, and encouraged a more regulated breathing pattern. The standing condition allowed subjects greater freedom to vary their breathing patterns, thus generating larger fluctuations.

The current algorithim was developed based on evaluating healthy lung biosignals in a controlled setting. Future development will need to consider the biosignals produced by unhealthy lungs, including sounds of different frequencies and amplitudes, such as crackles or asthmatic wheezes. Additionally, external acoustic noises such as talking or coughing that can take place in a hospital setting will need to be considered. To this end, a protocol is being submitted to Brigham and Women's Hospital's IRB.

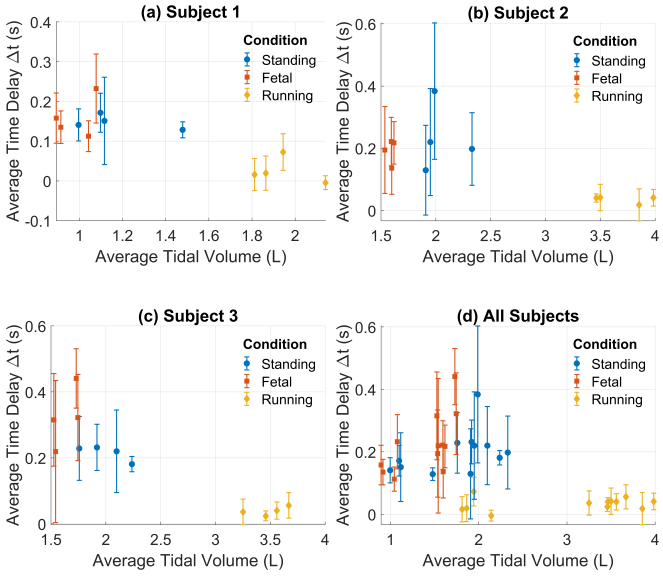

Fig. 5: Average time delay Δt plotted against average tidal volume for three healthy subjects and three subject conditions, as shown in (a)-(c). Four samples were collected for each subject condition. A comparison across subjects is shown in (d). The error bars represent one standard deviation from average Δt for each 30 second sample.

V. CONCLUSION

While this feasibility study comprised only a small number of healthy patients, the initial results are promising, indicating that Δt exists and that it changes with TV, as demonstrated by the difference between measurements conducted during and outside of physical activity. The sensing method is noninvasive and provides a quantitative basis to monitor lung condition over a period of time.

This method not only provides a baseline for future research, but with improvements, it has long-term potential to develop into an important tool in the clinical setting that provides an objective, quantitative metric for monitoring and assessing patients with restrictive respiratory conditions.

Near-term work includes hardware improvements to integrate all of the signals through one DAQ, which will improve signal synchronization, and a study with a larger data set of healthy patients to better understand respiratory mechanics, followed by testing on subjects with impaired lungs, to assess the utility of our developed method to monitor disease-induced changes. An open question, to be answered with a larger data set, is whether Δt provides a general indicator of TV or must always be baselined to a specific patient.

Longer term, the device would benefit from reduction to a wearable form factor which would reduce the risk of cross contamination and enable true continuous and remote patient monitoring.

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