According to the World Health Organisation (WHO), 235 million people suffer from respiratory illnesses which causes four million deaths annually. Regular lung health monitoring can lead to prognoses about deteriorating lung health conditions. This article presents our system SpiroMask that retrofits a microphone in consumer-grade masks (N95 and cloth masks) for continuous lung health monitoring. We evaluate our approach on 48 participants (including 14 with lung health issues) and find that we can estimate parameters such as lung volume and respiration rate within the approved error range by the American Thoracic Society (ATS). Further, we show that our approach is robust to sensor placement inside the mask.

CCS Concepts: • Human-centered computing → Ubiquitous and mobile computing design and evaluation methods;

Additional Key Words and Phrases: Pulmonary function test, wearable spirometry, smart mask

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1 INTRODUCTION
According to the World Health Organisation (WHO), 235 million people suffer from respiratory illnesses. Chronic Obstructive Pulmonary Disease (COPD) is associated with an estimated 3 million deaths per year, and asthma had an estimated 200 thousand deaths per year [55]. Common infectious respiratory diseases, such as influenza, cause 600 thousand deaths worldwide. 90% of deaths due to pulmonary disease occur in low-income and middle-income countries. COPD development generally starts early in life due to a complex interplay of disadvantageous factors, many of which occur in low- and middle-income countries [8]. Studies have shown

https://www.who.int/health-topics/chronic-respiratory-diseases.

The work was done while Tanmay Srivastava and Prerna Khanna were a research fellow at IIT Gandhinagar.

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that lung diagnosis is typically missed or delayed until poor lung health conditions advance. Early diagnosis of lung ailments can positively influence disease course, slowing progression, relieving symptoms, and reducing the incidence of exacerbation [65, 77]. But the unavailability of lung function equipment hinders proper diagnosis [70, 86].

Regular lung health monitoring can lead to prognoses about deteriorating lung health conditions [3, 74]. Typically, two kinds of breathing data are used to derive lung health bio-markers:

- **Tidal Breathing or Normal Breathing.** Tidal breathing refers to inhalation and exhalation during restful breathing. It is in involuntary action. Lung disease can change the normal characteristics of tidal breathing [48].
- **Forced Breathing.** It is performed by taking a deep inhalation resulting in full expansion of the chest followed by a forceful exhalation. A person will require to voluntarily perform forceful breathing maneuver.

**Forced Expiratory Flow (FEF)** measurement deduced from forced breathing is the most widely used method to assess the severity of asthma. Monitoring the forcefully exhaled airflow can help diagnose the onset of asthma, COPD and other conditions that affect breathing [37, 64, 76]. But, FEF measurements require a controlled environment. Moreover, a majority of young and old patients with airway obstruction are not able to perform adequate forced breathing maneuvers [76]. Prior research has shown that tidal breathing patterns can also be used to detect and quantify airway obstruction [80]. The respiration rate can be deduced from tidal breathing. It is defined as the number of breaths taken by a person in a minute. Respiratory rate deduced from tidal breathing is an important marker of cardiac arrest, dyspnea [17, 24, 34, 81], accessing sleep quality and monitoring stress [15]. Lung inflammation caused by COPD deterioration or lung infection leads to a higher respiration rate [37, 50].

In clinical settings and hospitals, exhaled airflow is measured using a spirometry test (Figure 3(a)). During a spirometry test, a patient performs forceful breathing through a flow-monitoring device (a tube or mouthpiece), which measures instantaneous flow and cumulative exhaled volume. However, spirometry tests performed at hospitals are not transient, and the recent global pandemic has led to the suspension of certain non-urgent healthcare services such as routine diagnostic testing [29, 33]. Although home spirometry tests are available [47], even the cheapest hand-held digital spirometer cost about USD 300. Previous work has accurately estimated forced breathing parameters using a smartphone [36]. In smartphone spirometry, a person must do a maneuver of forceful breathing towards the cellphone. But, the approach to microphone-based smartphone spirometry is susceptible to users’ way of holding the phone, users’ lip posture, environmental variability, and phone’s make and model [19].

Previous studies [50] have also used accelerometers in smartphones for continuous respiration monitoring. However, the approach requires a controlled environment where the participant must put the smartphone in a particular location on the chest. Recently, studies [91] have shown that wearable spirometry can be conducted using a pressure sensor inside a specialised mask meant for athletes. Progress on mask spirometry is limited because (i) the evaluations were done on healthy adults alone, ii) there was no continuous monitoring of respiration rate, iii) athlete training masks are relatively costly (40-50 USD) compared to consumer-grade masks (5-10 USD For N95 mask), iv) athlete training masks are meant to restrict the oxygen received by a person to create a high-altitude environment², as such they cannot be a replacement for generally used cloth or N95 mask. In Section 9, we qualitatively show that the general population would not prefer a specialised mask as a daily wearable.

This article shows that consumer-grade masks (N95 and cloth masks) can be used for Spirometry and continuous respiration rate monitoring by processing the signal from the microphone retrofitted inside the mask (Figure 1). The main intuition behind using audio is to leverage the relationship between variation in air flow rate and the intensity change in nasal sound [35, 43, 44, 87]. Our work addresses the limitations of prior research [1, 37, 91]. In particular, our approach to SpiroMask (i) provides a relatively controlled environment for

²https://www.healthline.com/health/training-mask-benefits#benefits.
Spirometry as well as tidal breathing monitoring, (ii) can classify between tidal breathing, noise, and speech, (iii) is accurate for participants with lung ailments as well as for healthy individuals.

Our approach uses the audio signal to derive vital lung parameters and continually monitor respiration rate. We used machine learning with sequential forward selection techniques to learn a set of audio features that accurately estimate lung health. We have two separate pipelines for estimating parameters from forced breathing and tidal breathing. To estimate respiration rate (from tidal breathing), we used a neural network that distinguishes tidal breathing from speech and noise. We estimated the average peak-to-peak time from the tidal breathing signal to derive the respiration rate. The parameters estimated from forced breathing are described in Section 2.

Our study was approved by the Institutional Review Board (IRB). We recruited 48 participants, including 15 female participants. A total of 14 participants had restrictive and obstructive lung ailment. The number of healthy and unhealthy participants in our trial is comparable to study population of past studies [20, 37, 52, 83, 88, 91, 92]. For each participant, the study lasted between 45 minutes to about an hour. The mean percentage errors on forceful breathing parameters for cloth masks were between 5.2% to 6.7%. For the N95 mask, they were between 5.8% to 6.3%. We achieved an accuracy of 94.7% on classifying tidal breathing from noise and speech. The Mean Absolute Error (MAE) on the estimation of respiration rate was 0.68 for the cloth mask and 0.49 for the N95 mask. Our results on forceful breathing are within the acceptable error range as endorsed by the American Thoracic Society or ATS.

We have performed sensitivity analysis on the position of the sensor inside the mask. Our approach is robust to sensor placement for forced breathing. However, certain positions (directly below the nose) inside the mask are ideal for estimating respiration rate from tidal breathing monitoring.

To summarise, the main contributions of this article are:

- **SpiroMask**: A novel mask-based system for estimating forced and tidal breathing to assess lung health parameters using a microphone
  - that works on consumer-grade masks;
  - is accurate within the ATS guidelines;
  - works well for both healthy and unhealthy subjects;
  - is robust to the microphone placement.
- **Public Dataset**: We publicly release our dataset at Github.\(^3\) We believe ours is the first such large publicly available dataset that measures both the tidal and forced breathing parameters and ground truth for 48 participants (including 14 with lung ailments). We believe that our dataset can help advance research in the community.
- **Reproducibility**: We believe that our work is fully reproducible. We use the same repository as above for the code. All the generated tables and graphs have corresponding scripts to reproduce all the results.

\(^3\)https://github.com/rishi-a/SpiroMask.
believe our efforts towards reproducibility will lower the effort towards replication and building on top of our work.

2 BACKGROUND AND RELATED WORK

This section underlines the basics of spirometry and lung function indices, followed by an overview of prior work on audio-based and pressure-based sensing of lung indices.

2.1 Spirometry

Spirometry is a widely used pulmonary function test. It measures how fast and how much air the patient can breathe out and is the most widely employed objective measure of lung function [75]. Spirometry tests are usually performed in clinics or hospitals. Currently, the most commonly used devices for respiratory evaluation are hand-held spirometers. A spirometry test consists of the following sequence of events [21]:

- A soft clip is placed on the patient’s nose to allow normal breathing through her mouth;
- The patient wraps her lips tightly around the spirometer mouthpiece as shown in Figure 3(a), ensuring that all the exhaled air goes through into the spirometer for accurate measurement;
- The patient then takes the deepest possible breath, filling her lungs to the maximum;
- The patient exhales hard and fast and continues exhaling into the spirometer until no more air comes out.

Spirometry results are effort dependent. Patients need to be coached by a trained professional to perform a forceful exhalation for a successful spirometry test. A digital spirometry test produces the flow versus volume plot of the lung from which vital lung parameters are extracted. A forceful breathing maneuver requires a controlled environment with proper guidance from doctors.

A standard spirometer measures the volume and flow of air that can be inhaled and exhaled. Spirometry generates pneumotachographs, which plot the volume and flow of air coming in and out of the lungs. The most common parameters measured by spirometry are **forced vital capacity (FVC)**, **forced expiratory volume at one second (FEV1)**, and **peak expiratory flow (PEF)**. FVC is the total air volume exhaled denoted by C in Table 1, FEV1 is the exhaled air volume in the first second of exhalation, denoted by B. PEF is the maximum airflow velocity in exhalation, denoted by A.

2.2 Smartphone Spirometry

Researchers have explored the potential to turn existing mobile devices and smartphones into portable electronic spirometers using their inbuilt microphones supplemented with machine learning techniques [36, 51, 63, 64, 73, 79, 92]. These studies use the features from the Hilbert transformation [28], linear predictive coding [46], and the spectrogram of an audio signal to train a linear regression model for each of the vital lung parameters, i.e., FVC, FEV1, and PEF. Compared to the clinical spirometer, the reported mean error is 5.1%, 5.2%, and 6.3% for FEV1, FVC, and PEF, respectively. In our work, we show that Hilbert transformation with a finite impulse response filter gives a comparable or better estimate of lung function parameters, and we can avoid non-trivial modelling like linear predictive coding. Also, given that the audio sensor is fixed in the mask, we are not required to compensate for pressure losses sustained over the variable distance from the mouth to the microphone, and reverberation/reflections caused in and around the subject’s body.

Researchers have also proposed a **variable frequency complex demodulation method (VFCDM)** technique to extract the FEV1/FVC ratio [73] from audio. A built-in smartphone microphone was also used in [92] to estimate FEV1 and FVC. However, they do not estimate PEF or tidal breathing parameters like respiration rate. Commodity smartphone has also been used to measure the humans’ chest wall motion via acoustic sensing. Lung function indices are deduced from the measured motion [64]. Existing literature on extracting PFT parameters

https://pftforum.com/review/tutorials/spirometry-tutorials/assessing-flow-volume-loops/.
Table 1. Forced Breathing and Tidal Breathing Lung Function Parameters

| ID | Abbreviation | Definition |
|----|--------------|------------|
| A  | PEF          | Peak expiratory flow (PEF) is the maximum airflow velocity in exhalation. (L/s) |
| B  | FEV1         | Forced expiratory volume in 1 second (FEV1) is the exhaled air volume in the first second of exhalation (L) |
| C  | FVC          | Forced vital capacity (FVC) is the total air volume exhaled (L) |
| D  | FEV1/FVC     | FEV1/FVC is the ratio of FEV1 and FVC. This ratio should be >80% among healthy participants |
| E  | RR           | Respiratory Rate (samples/min) |

Smartphone spirometry has the following limitations:

- It requires accounting for variability in the distance between microphone and mouth. In Section A.2, we have described how this variability affects the flow-volume curve;
- Not all smartphones are created equal. The flow detected by the microphone in smartphones relies on the mechanical transduction of sound, which is affected by the position of the microphone and the physical casing surrounding it;
- It is not suited for sensing tidal breathing as a person will not keep the smartphone near their nose for a long duration;
- It is relatively less accurate for participants with lung ailments;
- It requires a user to actively interact with the smartphone.

2.3 Wearable Spirometry

Recent efforts have been made to integrate face masks with audio and pressure sensors to extract vital lung parameters. Researchers have used a MEMS-based barometric pressure sensor inside an athlete training mask. Such masks are specially designed for athletes, restricting breathing, making an athlete feel like they are at a high altitude. Accurate wearable spirometry has been performed in athlete masks with error margins of 2.9% and 3.3% for FVC and FEV1, respectively. Researchers have also experimented by integrating an audio

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5https://www.trainingmask.com/training-masks/training-mask-3-0/ Last accessed: 15th August 2021.
sensor inside a surgical mask to estimate FVC from the energy of the audio signal but they have not quantified the error [1]. However, progress in mask spirometry is limited in the following ways:

- Continuous monitoring of tidal parameters was absent;
- The proposed approaches are not suitable for consumer-grade N95 and cloth masks;
- Previous work did not consider participants with lung ailments in their user-study and thus the efficacy is not known;
- There was no discussion on the robustness of sensor positioning inside the mask.

Table 2 compares our work with previous literature, and shows the novelty of our work for: (i) estimating lung parameters for consumer-grade cloth and N95 masks; (ii) single system for measuring both forced and tidal breathing parameters; and (iii) a larger study with unhealthy and healthy subjects; and a discussion on robustness of sensor placement.

### Table 2. Previous Studies on Mask-based Lung Function Measurement: (i) Do Not Work on Consumer-Grade Masks; (ii) Are Not Robust to Sensor Placement; (iii) Can Estimate Respiration Rate Only in Controlled Settings; and (iv) Did Not Consider Participants with Lung Ailments

| Type of mask | Participants | Lung health parameters | Discussion on sensor positioning inside the mask |
|--------------|--------------|------------------------|-------------------------------------------------|
| Cloth/Surgical/ N95 | Healthy | Unhealthy | Continuous monitoring of respiration rate | Forced breathing parameters |
| Zhou et al. [91] | ✗ | ✗ | 20 | 0 | ✗ | PEF, FEV1, FVC | ✗ |
| Adhikary et al. [1] | ✓ | ✗ | 8 | 0 | ✗ | FVC | ✗ |
| Curtiss et al. [14] | ✓ | ✓ | 9 | 0 | ✓ | ✓ | ✓ |
| SpiroMask (Our work) | ✓ | ✓ | 34 | 14 | ✓ | PEF, FEV1, FVC | ✓ |

2.4 Respiration Rate Detection

Most of the prior work estimates respiration rate using the Inertial Measurement Unit (IMU). Röddiger et al. [53] estimated respiration rate from an accelerometer and a gyroscope placed inside in-ear headphones. Hernandez et al. [22] installed IMU in Google glass to estimate pulse rate and respiration rate from a ballistocardiogram. Sun et al. [67] used accelerometer data from smartwatches to deduce respiration rate when a person is asleep. Liaqat et al. [38] used machine learning algorithms to filter out IMU data that can be used to estimate respiration rate in the wild using smartwatch. Aly et al. [4] used the approach of placing a smartphone over the chest to estimate respiration rate using an accelerometer and a gyroscope. Islam et al. [25] used sensor fusion data from both a smartwatch and a smartphone. All of these works except the work done by Sun et al. [67] focused on estimating respiration rate with a different posture of a user. For a fair comparison, we compared the error of only ‘Sitting’ posture with our work and found that our results are comparable or better to previous works. We have summarised many previous works in Table 3 below.

In the past, microphones have also been used to estimate respiration rate. Kumar et al. [35] used short audio segments from in-ear headphones to estimate respiration rate when a person is subject to strenuous exercises. Nam et al. [44] used a microphone placed below the person’s nose to record nasal sound. We cannot directly compare the error between SpiroMask and previous work that used a microphone due to different metrics used but we believe that SpiroMask performance is acceptable, given that our evaluation was on a relatively larger set of participants.
Table 3. SpiroMask’s Performance on Respiration Rate Detection Is Better or Comparable to Previous Work

| Metric | Error | Hardware (via Accelerometer) | Participants |
|--------|-------|------------------------------|--------------|
| Röddiger et al. [53] | MAE = 3.10 | Bluetooth earphone | 12 Participants |
|        | MAE = 2.74 | (via Gyroscope) | (2 Females) |
| Hernandez et al. [22] | MAE = 1.15 | Head mounted device | 12 Participants |
|        | MAE = 1.1 | (via Accelerometer) | (6 Females) |
|        | MAE = 1.1 | (via Camera) | |
| Kumar et al. [35] | MSE = 0.2 | Headphones | 21 Participants |
| Sun et al. [66] | MAE = 0.75 | Smartwatch | 16 Participants |
| Hernandez et al. [23] | MAE = 0.22 | Smartwatch | 12 Participants |
|        | MAE = 0.41 | (via Gyroscope) | (both) |
|        | MAE = 0.22 | (via Accelerometer) | |
| Liaqat et al. [38] | MAE = 1.9 | Smartwatch | 14 Participants |
| Islam et al. [25] | MAPE = 8.26% | Smartphone | 131 Participants |
| Aly et al. [4] | MdAE* = 0.04 | Smartphone | 7 participants |
|        | MdAE = 0.53 | (via Accelerometer) | (4 females) |
| SpiroMask (our work) | MAE = 0.6 | Consumer grade mask | 37 Participants |
|        | (13 Female) | (via microphone) | (14 with lung ailments) |

MAE is Mean Absolute Error and MdAE is Median Absolute Error.

2.5 Audio and Pressure Sensor

Prior literature has successfully used both an in-situ microphone and a MEMS barometer inside a very tightly sealed athlete mask to sense breathing and perform portable spirometry [1, 91]. However, our experiments show that a pressure sensor cannot be used to sense breathing inside a consumer-grade cloth mask. Figures 2(a) and 2(b) shows that the pressure sensor outputs a similar signal (Pearson correlation coefficient, $r = 88\%$) when the cloth mask is worn and not worn. The amplitude in Figure 2(b) is higher than in Figure 2(a) because the pressure inside a mask is higher than the atmospheric pressure. However, there is little to no change in the signal to distinguish between inhalation and exhalation. Previous work [91] leveraged the change in pressure to detect forceful breathing inside a special kind of mask. However, the differential pressure is insignificant in N95 and cloth masks making the breathing signal unrecognisable. It implies that pressure sensors are ineffective in distinguishing tidal breathing in cloth masks unless the sensor is sealed from atmospheric pressure. Figures 2(c) and 2(d) shows that a microphone does a better job (Pearson correlation coefficient, $r = 1\%$) in sensing tidal breathing inside a cloth mask. These results suggest that commodity-grade pressure sensors placed on standard masks are likely to result in poorer lung health estimation than audio sensing. Thus, we do not baseline against this previous work [91] in mask-based spirometry as the approach does not work on consumer-grade masks.

3 DATA COLLECTION PROCEDURE

We now describe our data collection procedure.

3.1 IRB Approval and Screening Criteria

Our user study on SpiroMask was approved by the Institutional Review Board (IRB). All participants between 18 to 70 years of age could become a participant in the study. People who are severely ill and have been advised...
Fig. 2. (a) and (b) shows that tidal breathing is indistinguishable from a pressure sensor placed inside the cloth mask whereas (c) and (d) show that a microphone is more suitable inside a cloth mask to monitor tidal breathing. Thus, previous work [91] in mask-based spirometry does not work on consumer-grade masks.

bed rest by doctors were not allowed to participate in the study. We recruited 52 participants for the study, out of which 16 participants reported having lung ailments. All participants were remunerated as per institute guidelines.

3.2 Entry Survey
Every participant filled out an entry survey before the user study. We asked them to declare their age, height, weight, any recent illness, history of contracting COVID19 and if they had a meal before coming for the user study. We also asked them if they had a clinically validated history of lung ailments and if they had done a spirometry test in the past.

3.3 Sensing Hardware
3.3.1 Microphone. We recorded the audio of forceful expiration and tidal breathing using an Arduino nano sense microcontroller due to the availability of embedded sensors like the MEMS microphone. Its compact size makes it possible to affix the microcontroller on a face mask. The microphone has a sampling rate of 16 kHz. We placed our microcontroller inside a 3D printed enclosure (Figure 3(f)) to protect it from static discharge and wear and tear. The 3D enclosure is affixed to the mask using velcro and double-sided tape (Figure 3(e)). We advised the participants to affix the microcontroller on their mask to ensure their comfort and safety while wearing it. We confirmed that the mask was appropriately worn before starting the SpiroMask experiment. Our future system would be smaller and hand-sewable on the fabric.

6https://store.arduino.cc/usa/nano-33-ble-sense.
Fig. 3. (a) and (b) shows the spirometry test followed by a SpiroMask test. (c) shows tidal breathing user-study where a smartphone is placed on the sternum with the help of a belt. (d) shows how the retrofit device was fixed into both the type of mask. (e) shows the protective velcro tape layer above the sensor to protect it from mucus, and (f) shows the microcontroller placed inside the 3D printed casing.

3.3.2 Spirometer. Our setup also consisted of the Helios 401 medical-grade (ISO 14971:2019) hand-held spirometer. We used the spirometer to collect the ground truth lung capacity of every participant.

3.3.3 Smartphone. We used a Samsung Galaxy M20 smartphone to collect accelerometer data for the expansion and contraction of the chest. The accelerometer had a sampling rate of 100 Hz.

3.4 Data Communication and Storage
The sensor inside the mask was connected to a desktop computer via USB cable. To ensure no data loss and interruption (due to draining battery) during the user study, we refrained from using any wireless mode of audio data transfer. Transmitting high-quality audio with limited power is a challenge because our sensing hardware samples audio at 16 kHz which amounts to 768 Kbps of data. But the practical throughput of BLE is around 125 Kbps or less. For every participant, the investigator checked if the microphone and the smartphone were responding to remote commands. We used the MATLAB mobile application to collect accelerometer data from the smartphone. An investigator could issue remote commands from their laptop to retrieve audio and IMU data from the microphone and smartphone. The sensitive audio data was stored securely in the cloud.

3.5 COVID19 Norms
We followed all local COVID19 guidelines during the entire process. The tests were done in a well-ventilated room with a single participant at a time. The investigator was double vaccinated and wore an N95 mask during the entire duration of the experiment. On the participant’s arrival, we asked the participant to sanitise their hand.

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7 https://labsakha.com/wp-content/uploads/2019/03/RMS-Helios-401-Features-Specifications.pdf.
8 https://www.mathworks.com/products/matlab-mobile.html.
Table 4. Demographic Information for the Participants

|                                | Forced Breathing | Tidal Breathing |
|--------------------------------|------------------|-----------------|
| Total Participants (n)         | 48               | 37              |
| Participants with lung ailments (n, %) | 14 (29.1%)       | 14 (37.8%)      |
| Females (n, %)                 | 15 (31.3%)       | 13 (35.1%)      |
| Age (yrs) (mean, range)        | 28.05 (21-68)    | 26.19 (20-32)   |
| Height (cm) (mean, range)      | 167.39 (142.4-182.9) | 167.0 (152.4-182.88) |
| Participants with lung ailments |                  |                 |
| Asthma                         | 3                |                 |
| Other Restrictive lung disorder | 4                |                 |
| Obstructive lung ailment       | 2                |                 |
| Post-COVID lung infection      | 4                |                 |
| Wheeze                         | 1                |                 |

hands. In the spirometry test, each participant used a new mouthpiece. We did not reuse any N95 or cloth masks. The investigator gave a fresh piece of both the masks.

3.6 Data Collection for Forced Breathing

We started the user study with data collection for forced breathing, as shown in Figure 3. We conducted the study over two phases. Phase 1 was in January - February 2021, and Phase 2 began in July 2021. Table 4 summarises the participant demography.

The investigators explained the entire user study to the participants. We demonstrated the spirometry test to the participants. For most of the participants, we could obtain the spirometry ground truth in the first attempt itself. The proprietary spirometer software flagged incorrect attempts for some participants. For every wrong maneuver, we repeated the spirometry test at least once, after which we obtained the ground truth. SpiroMask test followed the spirometry test. Forceful exhalation using the mouth is made possible in a hand-held spirometer due to the availability of a dedicated mouthpiece. Based on some pilot experiments, we realised that it is easier to do a forceful exhalation using the nose with the mouth closed when a face mask is worn. We asked the participants themselves to attach the sensor to the mask. We issued the remote command to start data collection and instructed the participant to begin deep inhalation and forceful exhalation. Each forceful breath lasted for 6 to 8 seconds. For 12 participants, we collected samples of forceful breathing audio for different positions inside the mask (L1, C1, R1, L3, R3 in Figure 4). Eight (66.6%) out of these 12 participants preferred to start with the R3 position. Most of the other participants, who did not participate in the sensor positioning study also preferred the R3 position. They all had an unanimous opinion that R3 was the most comfortable position to place the sensor as it did not lead to any discomfort. To ensure that forced breathing performance did not degrade in the course of multiple manoeuver, we asked all the 12 participants to rest for at least 5 minutes after each attempt. The ATS guideline permits up to 8 manoeuver in adults if the correct flow-volume curve is not achieved. The study duration for these subset of 12 participants was relatively longer compared to other participants. We then repeated the SpiroMask test using the cloth mask.

Despite several attempts, four participants (not included in the pool of 48 participants) were unable to perform the spirometry test. One participant had buccinator muscle pain making it impossible for her to hold the mouthpiece of the spirometer. Three other participants could not perform a proper forceful breathing manoeuver. They complained about lung discomfort when asked to attempt a forceful exhalation. The investigator decided not to continue with the spirometry test for these three individuals anticipating worsening medical conditions.

9Our country was very significantly impacted by COVID19 between March and June 2021 and thus we suspended data collection during that time.
It should be noted that while these three people expressed relative comfort using our system SpiroMask, we still had to discard their samples owing to the lack of verified ground truth.

We finally had data for 48 participants, out of which 14 had lung ailments. The number of healthy and unhealthy participants in our study is comparable or more than in several similar studies \[20, 37, 52, 83, 88, 91, 92]\.

3.7 Data Collection for Tidal Breathing

Data collection on tidal breathing started in the second phase (July 2021) of our user study. Table 4 summarises the participant demography. First, we asked the participants to place the smartphone in the sternum with the help of a chest belt, as shown in Figure 3. Previous work \[12\] has shown that putting an accelerometer in the sternum can help us retrieve the respiration rate by leveraging the movement of the rib cage and the movement of the abdomen. Accelerometer data are susceptible to motion artifacts, therefore we had positioned the participants on stationary chairs without any movable parts in it and advised them to remain as still as possible. To filter out noises due to involuntary movements, we used convolution filter. Convolution has the effect of a low pass filter \[16\]. We experimentally found out that using a window size of 70 samples denoises the signal. The accelerometer’s sampling rate was 100 Hz. Previous work has also used a stretch sensor combined with a motion sensor in a chest belt to measure breathing parameters, even when the user is ambulatory \[83\]. Proprietary respiration belts were used in some studies \[62, 88\] to collect the ground truth on respiration rate. Such ground truth systems are very expensive and unavailable in our country.

Similar to the SpiroMask test for forced breathing, we asked the participant to wear the mask and start tidal breathing. We issued remote commands from a laptop to start data collection from the smartphone and the microphone simultaneously. We also instructed the participants to begin counting their exhalations after issuing the remote command. Each sample of tidal breathing lasted for 20 seconds. We used a development platform \[27\] in which the time length of audio recording from the microcontroller was restricted to 20 seconds. The length of the audio recording could be extended had we used the Software Development Toolkit (SDK) of the microprocessor. We did not choose to do so due to engineering challenges. Prior work \[53\] has shown that 20 seconds of IMU data is enough to estimate respiration rate. In Appendix A, we have demonstrated why 20 seconds of audio is enough to estimate respiration rate. For 18 out of 37 participants, \[{10}\] we took at least two samples at each position (L1, C1, R1, L3, R3 in Figure 4) of the mask. These positions covered the entire space of

\[10\]We should recall that 48 users participated in the Forced breathing test and 37 users participated in the tidal breathing test as summarised in Table 4.
both the masks where the sensor could be placed. We repeated the SpiroMask test for tidal breathing on a cloth mask. Some participants were not comfortable in strapping the smartphone to the chest with a belt. For all such participants, we relied on their self-count and a metronome test.

We also performed a metronome test [58] for a subset of participants to validate the use of accelerometers to detect actual respiration rate. Previous studies have explicitly used a metronome as a ground truth to monitor respiration rate [12]. The metronome test is similar to the procedure of tidal breathing. But, in addition, we asked the participant to inhale and exhale as per the clicks of a 40 Beat Per Minute (BPM) metronome (40 beats corresponds to 20 exhalations and 20 inhalations). The metronome clicks were played on quietly so that they are not captured in the audio of tidal breathing. We choose 40 BPM because the participants were more comfortable at a lower breathing pace. Figure 5 shows a 20-second window of audio and IMU data for a participant performing breathing as per the metronome. We filtered the IMU data using a moving average filter with a window size of 20 data points. The average peak-to-peak time is 3.2 seconds which gives us 6.25 breathing cycles $\left(\frac{20\text{beats}}{60\text{s}} \times 20\text{s} = 6.66\text{beats}\right)$ which is close to the theoretical number of $\frac{20\text{beats}}{60\text{s}} \times 20\text{s} = 6.66\text{beats}$. Figure 5 validates the use of accelerometer as ground truth for respiration rate.

3.8 Exit Survey

The participants finally filled out an exit survey where they gave their feedback and opinion on the comfort of masks and spirometry tests. Particularly, we asked the following optional questions:

1. Can you compare the spirometer and SpiroMask test—which did you prefer and why?
2. Which mask (out of N95 and cloth masks) feels more comfortable to you and why?
3. Can you compare the SpiroMask and phone with chest belt for respiration rate monitoring—which did you prefer and why?

The entire user study took approximately an hour per participant.
4 APPROACH

The goal of SpiroMask is to estimate vital lung parameters and respiration rate using audio of breathing maneuver. Our work is inspired by the advancement of smartphone [36] and wearable spirometry [91]. The main intuition behind using audio is to leverage the relationship between variation in air flow rate and the intensity change in tracheal sound [87]. We now describe the pipeline for estimating the forced breathing and tidal breathing parameters.

4.1 Forced Breathing

A spirometry test comprises of forceful exhalation through a flow monitoring device that measures instantaneous flow and cumulative exhaled volume (Table 1). Similar to a spirometry test, a user is required to wear our smart mask, breathe in their full lung volume, and forcefully exhale. The entire pipeline of extracting forced breathing parameters is shown in Figure 6. We now explain the steps below:

(i) **Recording Audio.** The microphone inside the mask records the exhalation and sends the audio data to a computer.

(ii) **Normalising Amplitude.** The audio data is normalised between $-1$ and $1$ [71].

(iii) **Clipping Audio.** Keeping in terms with the ATS guidelines [13], we extract the part of the signal which represents a second before the start of forceful exhalation till the end of it. The start of forceful exhalation is detected using a threshold amplitude.

(iv) **Envelope Detection.** The audio signal’s envelope can be assumed to be a reasonable approximation of the flow rate because it is a measure of the overall signal power (or amplitude) at low frequency [36]. We obtain an estimation of the acoustic envelope of the forceful exhalation using Hilbert Transform (HT) [28]. This method has been employed in previous acoustical flow estimation studies [36, 71]. To validate HT’s estimated envelope for a signal with multiple harmonics, we generated a synthetic amplitude modulated

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signal using a message signal (the envelope) and a carrier wave \[49, 60, 66\]. Figure A.1 shows that the envelope estimated is a poor fit to the true envelope. We improve the estimated envelope using a finite impulse response (FIR) filter with Kaiser window \[49\]. Since an audio signal comprises multiple harmonics, we will need to process the HT envelope using an FIR filter with Kaiser window. The estimated lung parameters’ error margin would depend on stopband attenuation and transition width of the filter.

(v) **Approximate of ‘Volume-Time’ (VT) Curve.** The cumulative sum of flow rate over time gives us the approximate measure of the volume of the lung.

(vi) **Approximate of ‘Flow-Volume’ (FV) Curve.** At this stage we have the approximate ‘flow versus time’ curve and the approximate ‘volume versus time’ curve. These two results can be combined to derive the ‘flow versus volume’ curve.

(vii) **Future Generation.** We created two sets of features. The first set of features were from the audio waveform and the second set of features was from the FV curve. These features were input into a regression model to predict PEF, FEV1, and FVC.

As shown in Figure 6(f), we generated acoustic features with a window size of 30 ms and a step size of 15 ms for each audio sample. The window and the step size were based on prior literature \[37\]. The features are \textit{Mel Filter Bank Energy (MFE)}, \textit{mean and variance normalised Mel Frequency Cepstral Coefficient (MFCC MVN)}, the power spectrum, and the Mel Bands of spectogram. Besides these features, we also generated temporal features from the audio waveform and the FV curve \[6\]. Temporal features include information about entropy, peak-to-peak distance, the centroid of the signal and so on. Previous literature \[6\] lists all the temporal features and how they are computed for time series data. All the features are described in Table A.1 in the Appendix.

MFE features are known to distinguish speakers by modelling the shape of the vocal tract \[40, 69\]. Previous literature has used MFCC, temporal and statistical features to classify abnormal lung sound, and non-speech sounds \[10, 52, 56\]. As stated in the previous work \[37\], the features in the frequency domain can be assumed to be amplitudes excited by reflection in the vocal track and the mouth opening and therefore should be proportional to the flow rate.

(viii) **Machine Learning.** Given the extracted features and ground truth from spirometry, we train three supervised regression models: linear regression, \textit{random forest (RF)}, and \textit{support vector regression (SVR)} models to predict PEF, FEV1, and FVC from 260 features including acoustic and temporal features from each audio file. There are two main reasons for choosing these three algorithms: (i) Linear regression is easy to implement and thus would be best suited for edge devices with limited compute capabilities and (ii) RF and SVR can learn non-linear decision surfaces. They are also non-parametric algorithms (SVR with Radial Basis Function (RBF) is non-parametric) which means that their complexity grows as more data is made available.

Specific Features. To predict PEF, we used the cumulative sum of MFE, Log MFE, MFCC MVN, Power Spectrum and Melspectogram as an input to the regression model. To predict FEV1, we clipped the portion of the audio waveform after 1 second of the start of exhalation. These changes in features were inspired by previous literature \[37\]. For FVC, we used the features from entire audio waveform.

Feature Selection. We used the \textit{sequential forward selection technique (SFS)} \[26\] to reduce the feature space for faster computation and prevent overfitting.

### 4.2 Tidal Breathing

Our objective to to estimate respiration rate from tidal breathing. The first step in this process is to classify the audio samples as speech, tidal breathing, or noise. Thereafter, an algorithm calculated the average peak-to-peak time for every sample classified as tidal breathing to derive the respiration rate.
Fig. 7. (a) Breathing information is within 500 Hz. (b) Horizontal lines or harmonics signify speech information. A significant amount of speech information is lost if frequencies above 500 Hz are removed. We used a bandpass filter with a low cut-off frequency of 50 Hz and a high cut-off frequency of 500 Hz. The choice of low cut-off frequency was to reject the noise floor. The choice of high cut-off frequency was such that it conserves breathing information but rejects speech information.

Classification Task. The entire pipeline of estimating respiration rate is explained below.

- **Feature Generation.** Pulmonary nasal sounds lie in the frequency range of 20 Hz to 2,500 Hz whereas speech is located in the range of 300–4,000 Hz [10, 59]. For each audio sample, we used a Butterworth band pass filter with a low cut-off frequency of 50 Hz and high cut-off frequency of 500 Hz. We experimentally found out that these choices of frequencies conserved all the tidal breathing information as shown in Figure 7. The audio samples were sliced using a rectangular window as shown in Figure 8(b). We describe the choice
Fig. 9. (c) and (f) shows a successful and unsuccessful forced breathing manoeuver due to expiratory curvilinearity [42], respectively. (b) and (e) are the corresponding volume-time plot. (a) and (d) are the corresponding audio recording.

of window size and the size of offset between subsequent window in Section 5.3. For each window, we generate MFE features. MFE features isolate vocal track components which are crucial for estimating lung health [11, 37, 52].

- **Classification.** We used a 1D Convolutional Neural Network (1D CNN) to classify speech, tidal breathing and noise [85, 89]. We collected 40 minutes of tidal breathing data during our user study. To ensure a relatively balanced dataset, we also had 36 minutes of speech and 40 minutes of noise data. The details about noise data is described in the next section. 1D CNNs can be efficiently trained with a limited dataset of 1D signals where a temporal dependence exists between the values [31]. As shown in Figure 8, we used a 2-layer CNN with dropout.

- **Estimating Respiratory Rate.** Once we have identified and isolated tidal breathing audio, we used a respiratory rate detection algorithm to detect the respiratory rate for samples labelled as tidal breathing by the 1D CNN. We applied the Hilbert Transform envelope detection algorithm to the audio samples as described in Section 4.1. We used a peak detection algorithm [78] over the Hilbert envelope. The total length of the tidal signal divided by the average peak-to-peak time gives us the respiration rate.

5 EVALUATION

5.1 Experimental Setting for Forced Breathing

**Signal Processing.** The amplitude of the forced breathing audio recordings was normalised between −1 and 1, and clipped in time. Then, Hilbert transform was applied to deduce the approximate flow versus time curve by applying the pipeline explained in Section 4. The shape of the ‘flow-volume’ curve depends on the design of a minimum-order FIR filter. We need a minimum order filter to save on processing time for each sample and to ensure numerical stability [61, 84]. The transition width (\(w\)) and the stopband ripple (\(s\)) of a FIR filter decide the order of the filter. We used \(w = \frac{2}{n}\) as the transition width, (where \(n\) is the sampling rate which is 12 kHz) and \(s = -10\)dB as the stopband ripple to obtain the correct shape of flow-volume curve. Finally, one of the investigators visually analysed the flow-volume curve of every participant for its correct shape [32, 39]. In our future work, a machine learning model would classify correct and incorrect flow-volume curve based on the approach in the previous work [39]. Figures 9(c) and 9(f) shows a correct and incorrect shape of the flow-volume curve, respectively. While Figure 9(c) shows end expiratory curvilinearity [42], the same curvilinearity is missing.
Fig. 10. For **N95 mask**, the prediction error for participants with healthy and unhealthy lung condition is acceptable as per the ATS guidelines [42] for the Random Forest regression model.

in Figure 9(f). Beside curvilinearity we also relied on previous literature [39] to distinguish between correct and incorrect shape of flow-volume curve. The proper shape implies a successful forceful breathing manoeuver.

**Cross-Validation.** In both RF and SVR, we use a nested leave-one-out (subject) cross-validation strategy (LOOCV). The outer loop is used for predicting the lung parameters for a test participant, where all but that participant is used in the train set. The inner loop is used to fine-tune the hyper-parameters. LOOCV is preferred because it has far less bias compared to the validation set approach where we randomly split the dataset into train/test/validation or use K-Fold cross-validation. LOOCV tends not to overestimate the test error rate as much as the validation approach does [26].

**Metric.** We reported the mean percentage error among all participants for FEV1, FVC, and PEF. The percentage error is given by $|v_a - v_e| \times 100$ where $v_a$ is the ground truth value and $v_e$ is the estimated value. Percentage error is chosen as a metric because the guidelines on standardised spirometry [42] as well as subsequent research [36, 54] report the error between two spirometry efforts in percentages. Using the same error metric across Spirometry research helps us in a fair comparison of the performances.

**Tuning Hyperparameters.** The number of trees in RF span from 5 to 500 in steps of 10. The nodes in RF were expanded until leaves contain less than two samples. For SVR, the hyperparameter space consisted of the regularisation parameter and the type of kernel among linear, radial basis function and polynomial. We used grid search strategy to find the optimal hyperparameters. We discuss the results of forced breathing in Section 5.2.

5.2 Results for Forced Breathing

Our overall result for N95 mask is shown in Figure 10. The RF regression performs the best among participants with a healthy and unhealthy lung condition. Across all the participants, we have a **Mean Percentage Error (MPE)** of 6.30% on PEF, 5.82% on FEV1, and 5.98% on FVC for the N95 mask. To explain the direction of bias, Figure 11 shows the Bland-Altman [7] plot of each lung function measure. The vertical axis shows the difference between SpiroMask and Spirometer. The horizontal axis shows the mean value of the two methods. Measures taken for healthy participants are shown in blue dots and unhealthy participants in orange dots. Lines indicating the ±2σ are shown as dashed lines. From the plot, it can be seen that SpiroMask generalises well across healthy and unhealthy participants. On average, the ground truth Spirometer has slightly higher value for all the three parameters. Although, the PEF has higher variability compared to FEV1 and FVC, but the agreement holds true for PEF as well. For the cloth mask (Figure 12), we have an MPE of 6.71% on PEF, 5.25% on FEV1, and 5.67%
Fig. 11. Bland-Altman plot for the N95nMask. For FVC and FEV1, there is an agreement between Spirometer and SpiroMask for participants with healthy and unhealthy lung. The same agreement exists for PEF but with relatively higher variability.

Fig. 12. For Cloth Mask, the prediction errors for cloth mask is comparable to N95 mask. The standard deviation of FVC is higher compared to N95 mask. The prediction error across all participants is within the ATS guidelines [42] for the Random Forest regression model.

Fig. 13. On FVC. The Bland-Altman plot for cloth mask (Figure 13) shows a similar agreement between spirometer and SpiroMask. Our results for all the lung parameters fall within the clinically relevant range. As mentioned in previous work [37], a clinically relevant range is used because a participant cannot simultaneously use a spirometer and SpiroMask, so actual ground truth is unattainable. The limit of variability for a measure of lung function value should be within 7% over short duration as mandated by the ATS guidelines [42]. Our result on both the masks are well within the ATS guidelines for both healthy and unhealthy participants. Details of confounding factors and trends are described in Section 8.

Our result on PEF is better on the N95 mask. FEV1 and FVC are marginally better on cloth masks. In the future, we plan to study the comparative performance of the two masks using statistical tests.

The category of features selected using Sequential Forward Selection for PEF, FEV1, and FVC are listed in the Appendix A.
5.3 Experimental Setting for Tidal Breathing

5.3.1 Classification Task. We had 120 samples of tidal breathing across 37 participants. Speech data was available from 108 participants (36 minutes of data). There were 120 samples of noise data (40 minutes). The noise class consists of noise collected from different places like crowded cafeteria (30 samples), artificial traffic noise played via YouTube in a Bluetooth speaker in different volume levels (40 samples), white noise (30 samples generated from Audacity\footnote{https://www.audacityteam.org/}) and 20 samples collected from a garden (mostly involving children playing and bird chirping). We have the following set of hyperparameters: the window size, offset size and Fast Fourier Transform (FFT) length (The choice of the hyperparameter space is described in the Appendix). The window size cannot be bigger than the shortest audio sample, and the offset size can at maximum be as large as the window size. We used 6-fold cross-validation where the inner loop was used for tuning hyperparameters.

5.3.2 Estimating Respiration Rate. For all samples that were classified as tidal breathing, we applied Hilbert Transformation over the sample, followed by peak detection. Peak detection was also applied on the accelerometer signal. An algorithm compared the average peak-to-peak time between the ground truth accelerometer signal and the audio signal. The total length of the signal divided by the average peak-to-peak time gave us the respiration rate. The respiration rate was compared using Mean Absolute Error (MAE) for each participant. Our choice of error is in line with most of the previous literature listed in Table 3 in Section 2.

5.4 Results for Tidal Breathing

5.4.1 Classification Task. Our result in Figure 14 shows that we achieve an accuracy of 94.7%. The confusion matrix shows that the CNN misclassified 2% of noise samples and 7% speech samples as tidal breathing. As described in Section 4.2, we segmented each audio sample into smaller windows, and the classifier predicts the class of every window. The CNN assigns one class (Tidal Breathing, Noise, or Speech) to an audio sample if 90% of the segmented windows of that sample belong to that class. We have set a high threshold to ensure a fewer number of false positives. Figure 14 also shows the percentage of samples which the CNN could not classify.
higher threshold leads to better accuracy, but at the cost of some samples labelled as uncertain. A lower threshold would lead to higher misclassification. For example, the peak detection algorithm would return an inaccurate respiration rate if the CNN labels a speech sample as a tidal sample. A higher threshold lessens misclassification by segregating audio samples that cannot be classified into one class.

5.4.2 Estimating Respiration Rate. We achieved a MAE of 0.49 on the N95 mask. The MAE on the cloth mask was 0.68. Our results are comparable to previous work on estimating respiration rate using smartphone [50] and WiFi signals [88]. We estimated the respiration rate from the average peak-to-peak time. It must be noted that the Hilbert Transform envelope for samples incorrectly classified as tidal breathing samples had an amplitude of zero resulting in no peaks, and we thus reject such cycles.

6 SENSITIVITY ANALYSIS

6.1 Sensor Positioning

In this section, we analyse the robustness of our methods with respect to sensor placement inside the mask. The motivation to perform this experiment is that a person might retrofit the sensor inside any position inside the mask, and SpiroMask will be expected to monitor forced and tidal breathing without any drop in accuracy of lung parameters.

We described in Section 3 that we collected forced breathing and tidal breathing samples for some participant by placing the sensor in five different positions inside the mask (Figure 4). We had sensor position data for 12 participants (including 8 participants with unhealthy lungs). For tidal breathing, we had sensor positioning data for 18 participants. We used the model trained on audio samples collected by placing the sensor at position L1 and predicted on all other samples that belong to locations C1, R1, L3, and R3. We used the same cross-validation strategy as described in Section 5. Figure 15 shows the main results on 12 participants (Forced Breathing) and 18 participants (Tidal Breathing). For forced breathing, we observe that SpiroMask is robust to sensor placement where the MPE for all positions is below 7% (the ATS acceptable error). However, for tidal breathing, locations L3 and R3 have the lowest MAE. The low MAE on L3 and R3 can be attributed to the microphone being below the apex of the nose. None of the audio samples from L3 or R3 were misclassified during the classification task explained in Section 5. A detailed break-up of MPE and MAE for forced and tidal breathing among healthy and unhealthy participants is given in Figures A.5 and A.6. Note that the error metrics were chosen based on prior work on Smartphone spirometry and respiration rate estimation as explained in Sections 5.1 and 5.3.
Fig. 15. Forced breathing is robust to sensor placement, with relatively better performance at position L3 and R3. Tidal breathing can be accurately monitored at position L3 and R3.

Fig. 16. Speech becomes indistinguishable below 8 kHz [18]. The classification accuracy drops from 94.7% to 91.7% when sampling rate is decreased from 16 kHz to 8 kHz.

6.2 Sampling Rate

For continuous respiration rate monitoring, SpiroMask does not analyse speech signals and is concerned only with tidal breathing signals. The information content in tidal breathing lies between 50 Hz to 500 Hz.

Figure 16 shows the classification accuracy for different sampling rates. The classification accuracy remains above 80% even when the sampling rate is reduced to 1 KHz. The decrease in accuracy is attributed to misclassification of noise and speech samples as ‘uncertain’ samples at a lower sampling rate. Figure 17 shows that the mean absolute error of average peak-to-peak time (consequently the respiration rate) calculated on tidal samples increases on decreasing the sampling rate. Reducing the sampling rate to 1 KHz also aids in aiding privacy concerns as speeches are not intelligible when we sample audio at 1 KHz. To validate speech intelligibility, we asked three participants to hear two speech recordings. Each of these recordings were sampled at 2 KHz and 1 KHz. Each sample size is 16-bits long. None of the participants could decipher the speeches recorded at 1 KHz.
Fig. 17. Mean Absolute Error of respiration rate detection on tidal samples increases on decreasing the sampling rate. MAE is 0.6 at 16 kHz; it increases to 0.68 at 1 kHz.

Table 5. Current Consumption of the CNN Classifier

| Current Consumption (mA) | Battery Life With 11h/day of Operation |
|--------------------------|---------------------------------------|
| No Operation             | 0.96                                  |
| Sampling Classifier      | 1.6                                   |
|                          | 4.7                                   |
|                          | 13 Days                               |

With a 240 mAh battery, the classifier can run for 13 days.

One of the three participants could decipher one of the speech recordings sampled at 2 KHz. Speech spoken relatively slowly is easier to decipher at a lower sampling rate. Therefore, for preserving privacy, we can use lower sampling rates at the cost of a slight decrease in the accuracy of the respiration rate.

7 ENERGY CONSUMPTION OF CLASSIFIER

To study the battery life and the usability of our system, we arrived at an estimated power consumption of SpiroMask’s tidal breathing monitoring system. We evaluated the power consumption of the classifier running on the microcontroller. We used an oscilloscope to calculate the voltage drop across a 68Ω shunt resistor. We arrived at the value of the shunt resistor on the basis of the voltage resolution of the oscilloscope so that a small current draw (approximately mA) leads to a high voltage drop (approximately 100 mV) across the shunt. Consequently, we could arrive at the current consumption. The microcontroller first samples the audio, after which the model inference predicts whether the sampled audio represents ‘Noise’, ‘Tidal Breathing’, or ‘Speech’. The current consumption at different stages of the microcontroller can be seen in Table 5. With a duty cycle of one respiratory rate every minute, the average current consumption is 1.64 mA. To arrive at a battery life estimate, we assume 11 hrs/day of active respiration rate measurement with a 240-mAh coin cell battery. In this scenario, SpiroMask’s respiration rate system is expected to last for approximately 13 days on battery power alone.

8 CONFOUNDING FACTORS AND TRENDS

We performed 8-way multivariate analysis of variance (MANOVA) [82] tests to determine if other variables significantly contributed to the difference (in terms of percent error) between SpiroMask and the spirometer in estimating the forced breathing parameters FEV1, FVC, and PEF. MANOVA is more suitable than univariate analysis of variance (ANOVA) tests for forced breathing since we have three dependent variables. We performed separate MANOVA tests for N95 and cloth masks. We used height, weight, gender, age, whether the subject has performed spirometry tests before, whether the subject reported any lung ailments, whether the subject has the habit of smoking, and whether the subject had a meal before appearing for the test as the eight grouping
variables. The test for cloth mask shows that none of the grouping variables have a significant effect on the percent error between SpiroMask and spirometer measures (all $p$-values were greater than 0.05, the significance level of the test). The test for N95 mask suggests that height could be a significant variable ($p$-value $\approx 0.04$). Since this is only suggested by the test for N95 mask and not by the test for cloth mask, we need to investigate further to determine how significant the variable height is. We have provided scatter plots showing the variation of percent error for the forced breathing parameters with respect to height (for N95 mask) in Section A.6 of the Appendix. The presence of trend between FVC and FEV1 with height suggests incorporating such information whenever available will further improve our models. We leave the detailed ablation study with participant features like health and weight for the future.

We also performed a 7-way ANOVA tests to determine if other variables significantly contributed to the difference (in terms of percent error) between SpiroMask and a smartphone strapped to the chest in estimating the tidal breathing parameter (respiration rate). We performed separate ANOVA tests for N95 and cloth masks. We used height, weight, gender, age, whether the subject reported any lung ailments, whether the subject has the habit of smoking, and whether the subject had a meal before appearing for the test as the seven grouping variables. The tests for both N95 and cloth masks show that none of the grouping variables have a significant effect on the percent error between SpiroMask and smartphone measures (all $p$-values were greater than 0.05, the significance level of the test). Refer to Section A.6 in the Appendix for more detailed results of the MANOVA and ANOVA tests.

9 PARTICIPANT FEEDBACK

We collected feedback through an exit survey with three optional questions: whether they preferred a spirometer or SpiroMask (for measuring forced breathing parameters), whether they preferred SpiroMask or a phone strapped to the chest with a belt (for measuring tidal breathing parameters), and whether they preferred a cloth mask or an N95 mask.

9.1 SpiroMask and Spirometer

Around 58% of the responses preferred SpiroMask over a spirometer, while around 21% of the responses were equally comfortable with both methods. The main reasons cited by the responses preferring SpiroMask over a spirometer included: “mask was more comfortable”, “forced exhalation was easier to perform in a mask”, and “spirometer was hard to hold and operate, it is bulky”. Some of the older aged subjects commented that “it becomes hard to hold the spirometer in the mouth for a long time” (subject no. 16), and “mask is preferable as spirometer required breathing using mouthpiece which was a bit difficult” (subject no. 7). Interestingly, 75% preferred a cloth mask over an N95 mask, mainly based on comfort.

9.2 SpiroMask and Chest Belt

As mentioned in Section 4, we used the accelerometer sensor of a smartphone to collect ground truth on respiration rate. But, we could attain the ground truth via accelerometer only for a few participants. For the rest, we relied on self-counting as the ground truth. Among those who wore the smartphone-strapped belt, 91% preferred the SpiroMask over the phone to monitor tidal breathing during the user study. A participant said, “the chest belt seemed to restrict breathing,” and others voiced similar opinions. When asked by the investigator if they would like to wear a chest belt or SpiroMask as a part of their daily life, the opinion unanimously favoured SpiroMask.

10 LIMITATIONS AND FUTURE WORK

We now discuss some limitations of our work and proposed future work to address them.

- Human motions hinder breathing measurement. Prior research has shown that sensors worn on the chest can be used to measure breathing parameters during activities such as walking [83]. But, we found that
wearing a chest belt is not always comfortable and is susceptible to jittering activity. Currently, SpiroMask can classify noise, breathing audio, and speech, but does not attempt to extract a breathing signal from noisy audio recordings. We investigated audio signals which were collected from the microphone placed inside a user donned mask in simultaneous presence of background noise and user activity. In our follow-up work [2], we demonstrated the challenges in extracting useful breathing information from such signals.

- Our confounding variable analysis suggests that incorporating personal parameters such as height can help improve the estimation for forced breathing parameters. In the future, we plan to conduct an ablation study to study the potential improvements in modelling when personal health parameters are available. Previous smartphone spirometry work has shown improvement via personalisation [36], though their notion of personalisation was different.
- A detailed large-scale user study on the usability of SpiroMask is currently out of the scope of our current research. In our current work, we evaluated the percentage of participants who would prefer SpiroMask over a traditional spirometer or a chest belt.
- Currently, SpiroMask could not detect inhalations due to low amplitude. This is a known problem in smartphone spirometry [36]. However, since we use a custom microphone instead of a smartphone, we plan to experiment with a multi-microphone setup; one with high gain for inhalation and the other for exhalation.
- In our current work, we do not estimate the tidal volume. This is primarily due to the lack of a clinical-grade ground truth device for tidal volume estimation. We believe that our existing pipeline for forced volume should be easily adaptable to estimate tidal volume.
- In our current work, we predict specific points on the flow-volume curve (such as FEV1, FVC). In the future, we plan to predict the whole flow-volume curve instead of only these points. This problem can be naturally mapped to a sequence-to-sequence learning problem [68]. We plan to leverage recent advances in neural-networks-based sequence to sequence methods for this task [45, 90].
- Our current prototype requires external power. Recent advances in the community have leveraged a triboelectric nanogenerator to develop a self-powered acoustic sensor [5]. In the future, we would like to explore these recent advances towards making the system self-powered.
- Our work shows that lung health can be diagnosed by monitoring forceful breathing or tidal breathing via a microphone placed inside a consumer-grade N95 or cloth mask. We optimised the CNN classifier described in Section 4.2 to work in real-time in the Arduino Nano 33 BLE Sense microcontroller. We used offline algorithms to deduce forced and tidal breathing parameters. We envision that a SpiroMask will feed the lung health parameters on a wearer’s smartphone over Bluetooth Low Energy (BLE).

11 CONCLUSION

In this article, we presented a system for performing spirometry and continuous respiration rate monitoring using consumer-grade N95 and cloth masks. Forced and tidal breathing are used for deriving lung health biomarkers like estimating the respiration rate and volume of exhaled air. We showed that a retrofit sensor placed inside an N95 or a two-layered cloth mask can estimate forced and tidal breathing. Our evaluation of over 48 participants of forced breathing implies that the accuracy of wearable spirometry is well within the clinically accepted range for participants with and without lung ailments. Moreover, our work is comparable to existing research on portable spirometry and requires less complex modelling, making it possible to deploy it on microcontrollers running machine learning models. Our subjective evaluation in our study population shows acceptability and ease of use compared to traditional spirometry.

\[12\]https://store-usa.arduino.cc/products/arduino-nano-33-ble-sense.
A APPENDICES

In this supplementary section, we provide additional details for our article.

A.1 Validating Hilbert Transformation

Figure A.1 below validates the Hilbert-Transform envelope detection algorithm for a synthetic signal with multiple harmonics.

Fig. A.1. (a) A poor and (b) a proper fit of the signal envelope using Hilbert transform without and with FIR filter respectively.

A.2 Comparison with Smartphone Spirometry

To compare SpiroMask and smartphone spirometry, we asked three participants to perform four forceful breathing manoeuvers. Participants performed the first two maneuvers using a smartphone. They used SpiroMask for the following two manoeuvers. Our objective was to quantify the difference between the two manoeuvers recorded via both systems. Figure A.2 shows the comparison for one such user. The spectrogram of two manoeuvers performed over SpiroMask are more similar (Figure A.2(a)) than those conducted over the phone (Figure A.2(b)). Particularly, the information between 2-6 kHz in one of the manoeuvers performed via smartphone is lost which leads to incorrect flow-volume curve [20]. The variability arises in the phone because a user is unable to repeat the maneuver with the same angle and the distance between the mouth and the phone. A dissimilar flow-volume (FV) curve for the same user over a short period (< 5 Minutes) leads to a wrong lung health estimate [54]. This experiment shows that SpiroMask provides a more controlled environment for spirometry. Figure A.2(a) also shows that the inhalation phase of breathing.

Fig. A.2. SpiroMask provides a more controlled environment to perform spirometry as compared to smartphone.

(a) The information content in the spectrogram is similar for both the maneuver performed in SpiroMask. Here, both the inhalation followed by exhalation can be seen (b) The information between 2-6KHz is missing in the first maneuver performed via smartphone.
A.3 Feature Set
The list of features are described in Table A.1.

| Group                | Frame Level Descriptors                                                                                                                                 |
|----------------------|--------------------------------------------------------------------------------------------------------------------------------------------------------|
| 1 Temporal Domain    | Autocorrelation, Centroid, Mean absolute difference, Mean difference Medlan absolute difference, Median difference, Distance, Sum of absolute difference, Total energy, Entropy, Peak-to-peak distance, Area under the curve, Absolute energy, Maximum peaks, Minimum peaks, Slope, Zero crossing rate. |
| 2 MFCC                | 13 Mel-Frequency Cepstral Coefficients                                                                                                                                 |
| 3 MFCC (Normalised)  | 13 Variance Normalised Mel-Frequency Cepstral Coefficients                                                                                                                                 |
| 4 Mel Spectogram      | 64 Mel Bands of Spectrogram                                                                                                                                 |
| 5 Power Spectrum      | Power spectrum for each frame for each audio waveform                                                                                                                                 |
| 6 Mel Filter Bank Energy | 40 Mel Bands of spectrogram                                                                                                                                 |
| 7 Log Mel Filter Bank Energy | 40 Mel Bands of spectrogram                                                                                                                                 |

Note that the temporal domain features were also computed for the Flow-Volume curve.

A.4 Length of Audio Signal for Tidal Breathing
In Section 3.7, we mentioned that we used 20 seconds of audio recording to estimate the respiration rate. We were limited to 20 seconds because of engineering challenges in our prototyping platform. To validate if 20 seconds of audio is enough to estimate respiration rate, we compared the 20-second recording with a 30-second recording collected from an alternate microphone. Both the microphones were placed inside the mask. One of these microphone was the onboard microphone of SpiroMask’s prototyping platform\(^\text{13}\) (Arduino’s microphone) while the other microphone was embedded in a breakout board which was connected to the Raspberry Pi Pico (RPi Pico Microphone).\(^\text{14}\) We ensured that both the microphone have the same make and model. The RPi Pico with an interfaced microphone could be identified as a USB device in a Linux computer. A participant donned the two-microphone mask and continued tidal breathing.

![Fig. A.3. The average time peak to peak between both signals is same. A 20-second audio can thus be used to estimate respiration rate.](image)

We simultaneously started recording breathing audio from both devices. The RPi Pico microphone could record audio for a long time since it was acting as a USB microphone connected to a computer. The envelope of both recordings are shown in Figure A.3. The Arduino microphone recorded a 20-second audio while we stopped the other recording at the end of 30 seconds. The average time between peak to peak for Arduino’s 20-second recording was 2.69s while for the RPi Pico’s 30-second recording was 2.72s. Both of the signals had 7 peaks at

\(^{13}\)https://docs.arduino.cc/hardware/nano-33-ble-sense/
\(^{14}\)https://www.raspberrypi.com/documentation/microcontrollers/raspberry-pi-pico.html.
the end of 20 seconds. The breathing rate estimated from the 20-second recording and the 30-second recording will be same. Thus, we conclude that 20 seconds of audio would suffice to estimate the respiration rate.

A.5 Ablation Studies

Tables A.2, A.3, and A.4 show the most important features for each of the three lung parameter (PEF, FEV1, and FVC) discovered using the Sequential Forward Selection (SFS) technique.

| Table A.2. Mean Absolute Percentage Error for Features Selected in Each Category using SFS for PEF in Cloth Mask |
|-------------------------------------------------------------|
| **Category of Feature For PEF**                              | **Mean Percentage Error** |
| 1 Temporal features of the FV curve                          | 13.71%                   |
| 2 Mel Filter Bank Energy (MFE) features                      | 13.32%                   |
| 3 Mean and variance normalised MFCC features                 | 9.92%                    |
| 4 Melspectrogram features                                   | 10.12%                   |
| 4 MFCC+MFE+Melspectrogram                                   | 6.89%                    |
| 5 All Features (by restricting audio waveform till the peak of the signal) | 6.71%                    |

The combination of thee features, i.e. MFCC, MFE and Melspectrogram gives comparable result.

| Table A.3. Mean Absolute Percentage Error for Features Selected in Each Category using SFS for FEV1 in Cloth Mask |
|-------------------------------------------------------------|
| **Category of Feature (FEV1)**                              | **Mean Percentage Error** |
| 2 Temporal features of the FV curve                          | 19.25%                   |
| 3 Mel Filter Bank Energy (MFE) features                      | 11.58%                   |
| 4 Log of Mel Filter Bank Energy (MFE) features               | 10.18%                   |
| 5 Mean and variance normalised MFCC features                 | 6.21%                    |
| 9 All Features (by restricting audio envelope till 1 second after the start of exhalation) | 5.25%                    |

MFCC alone achieves a error of 6.21% which is acceptable as per ATS criteria.

| Table A.4. Mean Absolute Percentage Error for Features Selected in Each Category using SFS for FVC in Cloth Mask |
|-------------------------------------------------------------|
| **Category of Feature (FVC)**                              | **Mean Percentage Error** |
| 1 Temporal features of the FV curve                          | 21.36%                   |
| 2 Mel Filter Bank Energy (MFE) features                      | 12.71%                   |
| 3 Log of Mel Filter Bank Energy (MFE) features               | 12.38%                   |
| 4 Mean and variance normalised MFCC features                 | 13.11%                   |
| 5 Power spectrum features                                   | 9.69%                    |
| 6 Melspectrogram features+MFCC                               | 6.16%                    |
| 7 All Features                                              | 5.67%                    |

It can be seen that melspectrogram in combination with MFCC gives comparable result.
A.6 Multivariate Analysis of Variance (MANOVA) Test

Tables A.5 and A.6 show the results obtained in the 8-way MANOVA and 7-way ANOVA tests respectively. MANOVA was performed for forced breathing parameters and ANOVA was performed for the tidal breathing parameter (respiration rate). Both MANOVA and ANOVA were performed for cloth and N95 masks. The eight grouping variables used in the MANOVA tests were height, weight, gender, age, whether the subject has performed spirometry tests before, whether the subject reported any lung ailments, whether the subject has a habit of smoking, and whether the subject had a meal before appearing for the experiment. The seven grouping variables used in the ANOVA tests were height, weight, gender, age, whether the subject reported any lung ailments, whether the subject has a habit of smoking, and whether the subject had a meal before appearing for the experiment. Figure A.4 presents the variation of percent error for the forced breathing parameters with respect to height (the grouping variable with the lowest $p$-value in the MANOVA test) for N95 mask.

### Table A.5. Results of the MANOVA Test for Both Cloth and N95 Masks Are Shown

| Grouping Variable                        | p-value | Cloth Mask | N95 Mask |
|------------------------------------------|---------|------------|----------|
| Age                                      | 0.9886  | 0.5533     |          |
| Gender                                   | 0.8378  | 0.6739     |          |
| Height                                   | 0.9365  | 0.0383     |          |
| Weight                                   | 0.4568  | 0.5243     |          |
| Whether the subject had a meal before the experiment | 0.5802  | 0.1457     |          |
| Whether the subject has a habit of smoking | 0.8733  | 0.2166     |          |
| Whether the subject has performed spirometry before | 0.1021  | 0.7105     |          |
| Whether the subject reported any lung ailment | 0.0930  | 0.4712     |          |

The test for cloth mask shows that none of the grouping variables have a significant effect on the percent error between SpiroMask and spirometer measures (all $p$-values were greater than 0.05, the significance level of the test). The test for N95 mask suggests that height could be a significant variable ($p$-value $\approx 0.04$).

### Table A.6. Results of the ANOVA Test for Both Cloth and N95 Masks Are Shown

| Grouping Variable                        | p-value  | Cloth Mask | N95 Mask |
|------------------------------------------|----------|------------|----------|
| Age                                      | 0.9436   | 0.1311     |          |
| Gender                                   | 0.6492   | 0.7258     |          |
| Height                                   | 0.0796   | 0.7958     |          |
| Weight                                   | 0.4298   | 0.5990     |          |
| Whether the subject had a meal before the experiment | 0.2749   | 0.0881     |          |
| Whether the subject has a habit of smoking | 0.3807   | 0.8628     |          |
| Whether the subject reported any lung ailment | 0.4152   | 0.8152     |          |

None of the grouping variables have a significant effect on the percent error between SpiroMask and smartphone measures (all $p$-values are greater than 0.05, the significance level of the test).
A.7 Sensitivity Analysis on Position of Sensor

Figures A.5 and A.6 present a detailed break up of MPE and MAE for forced and tidal breathing parameters among healthy and unhealthy participants, respectively.

| N95 Mask | PEF | FEV1 | FVC | RR |
|----------|-----|------|-----|----|
| A        |     |      |     |    |
| L1       | 3.49% | 5.11% | 7.61% |    |
| C1       |     |      |      |    |
| R1       |     |      |      |    |
| L2       | 3.70% | 5.58% |      |    |
| C2       |     |      |      |    |
| R2       |     |      |      |    |
| L3       |     |      |      |    |
| C3       |     |      |      |    |
| R3       |     |      |      |    |
| B        |     |      |      |    |
| L1       | 6.40% | 3.51% | 6.59% |    |
| C1       |     |      |      |    |
| R1       |     |      |      |    |
| L2       | 2.88% | 4.12% |      |    |
| C2       |     |      |      |    |
| R2       |     |      |      |    |
| L3       |     |      |      |    |
| C3       |     |      |      |    |
| R3       |     |      |      |    |
| C        |     |      |      |    |
| L1       | 5.26% | 5.69% | 7.85% |    |
| C1       |     |      |      |    |
| R1       |     |      |      |    |
| L2       | 2.30% | 4.43% |      |    |
| C2       |     |      |      |    |
| R2       |     |      |      |    |
| L3       |     |      |      |    |
| C3       |     |      |      |    |
| R3       |     |      |      |    |
| D        |     |      |      |    |
| L1       | 0.42% | 0.44% | 0.45% |    |
| C1       |     |      |      |    |
| R1       |     |      |      |    |
| L2       | 0.05% | 0.04% |      |    |
| C2       |     |      |      |    |
| R2       |     |      |      |    |
| L3       |     |      |      |    |
| C3       |     |      |      |    |
| R3       |     |      |      |    |

| Cloth Mask | PEF | FEV1 | FVC | RR |
|-------------|-----|------|-----|----|
| E           |     |      |     |    |
| L1          | 8.59% | 3.32% | 7.61% |    |
| C1          |     |      |      |    |
| R1          |     |      |      |    |
| L2          | 2.66% | 1.95% |      |    |
| C2          |     |      |      |    |
| R2          |     |      |      |    |
| L3          |     |      |      |    |
| C3          |     |      |      |    |
| R3          |     |      |      |    |
| F           |     |      |      |    |
| L1          | 5.47% | 4.39% | 2.21% |    |
| C1          |     |      |      |    |
| R1          |     |      |      |    |
| L2          | 2.87% | 1.02% |      |    |
| C2          |     |      |      |    |
| R2          |     |      |      |    |
| L3          |     |      |      |    |
| C3          |     |      |      |    |
| R3          |     |      |      |    |
| G           |     |      |      |    |
| L1          | 7.39% | 5.70% | 7.47% |    |
| C1          |     |      |      |    |
| R1          |     |      |      |    |
| L2          | 3.86% | 0.91% |      |    |
| C2          |     |      |      |    |
| R2          |     |      |      |    |
| L3          |     |      |      |    |
| C3          |     |      |      |    |
| R3          |     |      |      |    |
| H           |     |      |      |    |
| L1          | 0.50% | 0.50% | 0.40% |    |
| C1          |     |      |      |    |
| R1          |     |      |      |    |
| L2          | 0.03% | 0.05% |      |    |
| C2          |     |      |      |    |
| R2          |     |      |      |    |
| L3          |     |      |      |    |
| C3          |     |      |      |    |
| R3          |     |      |      |    |
Fig. A.6. Errors in forced breathing parameters are presented for eight unhealthy subjects (for both types of masks) and errors in tidal breathing parameter (respiration rate) are presented for eight unhealthy subjects (for both types of masks).

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