Portable Forced Oscillation Device for Point-of-care Pulmonary Function Testing*

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***Abstract*— The forced oscillation technique (FOT) provides a simple and accurate approach for pulmonary function testing. However, most current devices are large and high cost, hence the test remains sparingly used. To address this problem, we verified the feasibility of a smart and portable forced oscillation device based on a small subwoofer and ultrasonic sensor that is targeted at point-of-care pulmonary function testing. In this paper, we first develop and optimize the signal processing algorithm for impedance estimation. Then we characterize the signal quality of programmable oscillatory waveforms with varying frequency, amplitude and duration. Finally, we evaluate the performance of the device against both mechanical models and human subjects. The results show that the coherence function is above 0.9 for all frequencies and the measurement error is less than 10%. The device yields good repeatability and satisfies the clinical diagnostic requirements.**

I. Introduction

Respiratory diseases have become a major health threat and affect nearly 10% of world's population [1]. To provide simple and robust lung function tests, a variety of methods have been developed [2]. Among them, spirometry is the gold standard for measuring pulmonary functions. However, it provides no structural information and is not suitable to several patient categories such as young children, senior subjects due to cooperation difficulty [3] [4]. As an alternative, the forced oscillation technique (FOT) minimizes patient cooperation by measuring the passive response of the respiratory system to external pressure oscillations [5][6][7]. However, currently available FOT devices are bulky and high cost, and hence not often used.

In this paper, we investigate if a portable device based on a small subwoofer and ultrasonic sensor could meet the FOT guidelines [8], and provide simple and accurate forced oscillation tests. To verify the feasibility of the device, we first optimized the impedance estimation algorithm for the portable device for the case with low SNR. Then, we programmed the device to generate varying oscillatory pressure waveforms and studied the influence of measurement duration, pressure amplitude, and frequency on the output signal quality. Finally, we evaluated the accuracy of the device by measuring 2 mechanical models with calibrated resistive load and assessed the repeatability of the device by conducting multiple tests with 6 human subjects. In this work, we approached the problem from a signal design perspective, and compensated for the low SNR of the portable device with specially designed oscillatory waveforms to ensure reliable measurement results.

In the past, several studies have been conducted to study the feasibility of portable FOT devices [9][10][11][12][13]. For example, a portable device based on a small speaker and microcontroller has been developed and is able to measure lung resistance at 5 Hz [9]. Another FOT device based on the piezoelectric actuators has also been reported and indicated the development of light weighted single frequency FOT device is feasible [13]. However, past works only measure the respiratory impedance value at a single frequency and thus do not provide sufficient diagnostic information for all lung conditions. Moreover, both devices use pneumotachometer for flow sensing, which results in considerable signal attenuation. In this work, we systematically studied the signal quality of the portable device and designed programmable input signals to accurately measure respiratory impedance at multiple frequencies. The healthy human subjects in this study are aged between 24 and 45 years old with no smoking history. The experimental procedures involving human subjects described in this paper were approved by the Institutional Review Board (IRB No. 825430-1). The results indicate that the device meets the FOT guidelines and is able to provide sufficient diagnostic information [8].

II. Basic principle of FOT

The forced oscillation technique determines respiratory mechanical parameters by superimposing external pressure oscillations on spontaneous breathing and measuring the resultant flow. The respiratory impedance value Z_{rs} is estimated as the complex ratio between oscillatory pressure P_{rs} and flow V_{rs} with respect to frequency f, and the diagnostic decisions are made based on the respiratory resistance R_{rs} and reactance X_{rs} as follows [5]:

$$Z_{rs}(f)=\frac{P_{rs}(f)}{V_{rs}(f)}=R_{rs}(f)+iX_{rs}(f), f\in[f_{min}, f_{max}], \quad (1)$$

where R_{rs} corresponds to the frictional force exerted on airflow by the airway, and X_{rs} represents the elastic and the inertial components of respiratory system [5]. Fig. 1 shows the representative tracings of lung resistance R_{rs} and reactance X_{rs} from a previous study [14].

Based on the physical properties of respiratory system,

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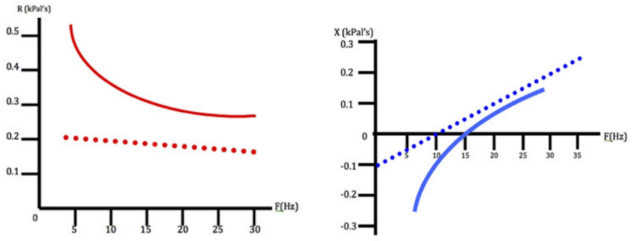


(a) (b)

Fig. 1. Representative tracings of lung resistance R_{rs}(a) and reactance X_{rs}(b). Solid lines: prototypical patients with distal obstruction; dotted lines: normal tracing. R_{rs} is elevated at $5Hz$ while the X_{rs} curve is right shifted when distal obstruction occurs (Figures adapted from [14]).

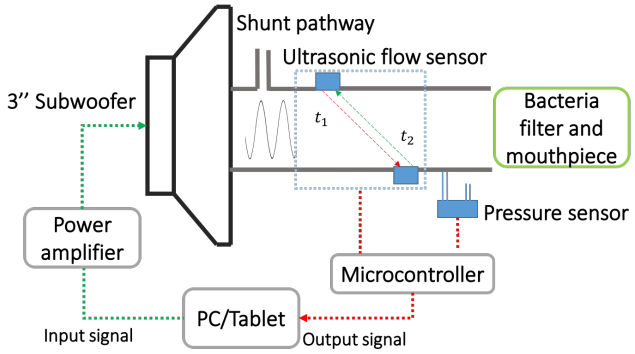


Fig. 2. Schematic representation of the prototype. t_1, t_2: transit time of downstream and upstream ultrasonic pulses.

the most informative frequency range is 5 Hz to 20 Hz [8]. Low-frequency pressure waves travel into lung periphery and provide information of the entire pulmonary system, whereas high-frequency oscillations only reach the proximal airway and inform about central airway conditions [5]. Thus, the device should be able to generate oscillatory pressure waves that cover a wide frequency range to satisfy the requirements of various diagnostic and monitoring purposes. In practice, the pressure oscillations are usually generated by a subwoofer [14]. The subwoofer size increases considerably in order to generate low-frequency pressure waves. Hence, the main challenges for a portable device are to achieve sufficient SNR and to detect small signals in noisy background.

III. Methods

A. Hardware design

The hardware design of the portable forced oscillation device mainly consists of two parts, which are: (1) oscillatory waveform generation, and (2) pressure and flow sensing. Fig. 2 shows the schematic representation of the prototype.

The input signal of user-selected waveforms is generated by the sound card of a PC/tablet and amplified by a Class-D power amplifier to drive a 3-inch subwoofer. The subwoofer generates oscillatory pressure waves at the airway opening. A shunt pathway is left open close to the speaker to enable spontaneous breathing of the subject. The pressure amplitude and flow rate at airway opening are measured by a 5.1cmH2O differential pressure sensor, and an ultrasonic flow sensor supplied by the Cognita Lab, LLC, which uses non-invasive ultrasonic sensing to minimize signal attenuation. The output signals are sampled at 500 Hz by a low-power microcontroller to automatically compute respiratory impedance value and the result is transmitted back to PC/tablet for visualization and storage. The prototype has a dimension of 8cm×8cm×18cm, a weight of 1.5 kg and could be powered through USB cable, thus portable and flexible for various applications.

The device supports the generation of mono- or multifrequency sinusoids between 5 Hz and 30 Hz. The frequency of the oscillatory waveforms could be programmed through the user-interface to satisfy various diagnostic or monitoring purposes. The power of each frequency is tuned to generate pressure oscillations between 0.1-0.3 kPa at airway opening, that guarantees sufficient SNR while minimizing patient discomfort. The duration of each test is 30-40 sec.

B. Impedance estimation algorithm

The pulmonary function test is done during spontaneous breathing. As a result, the outputs from pressure and flow sensor include both high-frequency excitation signals and low-frequency breathing components. By converting the entire signal into the frequency domain, we observe that breathing noise and its higher harmonics are mainly below 2 Hz, whereas the excitation signals are greater than or equal to 5 Hz. Thus, we use a 3rd order Butterworth filter to separate the excitation signal from the breathing noise [15]. The cutoff frequency of the filter is dynamically chosen depending on the frequency spectrum of breathing noise. Since the portable device has relatively lower SNR compared to large commercial devices, we estimate the respiratory impedance using cross- and auto-spectrum rather than directly compute the ratio of FFT to improve the estimator's performance against noise [14]. The estimator is shown in (2)

$$\widehat{Z}(f) = \frac{P(f) \cdot V^*(f)}{V(f) \cdot V^*(f)} = \frac{G_{PV}(f)}{G_{VV}(f)}, \tag{2}$$

where G_{PV} is the cross-power spectrum between oscillatory pressure and flow, and G_{VV} is the auto-power spectrum of flow. G_{PV}, G_{VV} are computed using Welch's averaged periodogram method [16]. The window length and window overlapping are set to 1000 and 500 data points respectively to minimize the estimation error.

C. Device calibration

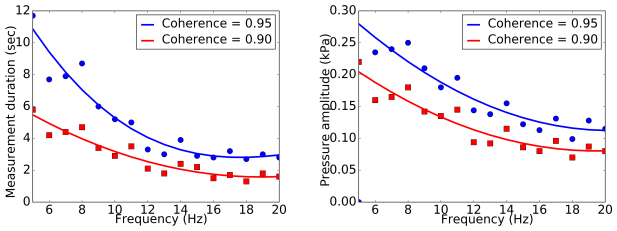
Since the pressure sensor is already temperature calibrated and has $\pm 0.25\%$ accuracy, we adopted a two-step process for device calibration. First we calibrated the ultrasonic

TABLE I. Maximum error of PEF (%)

	0.5(L/s)	1.0(L/s)	1.5(L/s)	2.0(L/s)
0.25Hz	6.69	4.04	1.71	3.03
0.32Hz	5.71	3.67	2.75	3.94
0.50Hz	5.07	3.62	2.98	3.83

TABLE II. Maximum error of PVC (%)

	0.5(L/s)	1.0(L/s)	1.5(L/s)	2.0(L/s)
0.25Hz	0.29	1.28	2.95	1.03
0.32Hz	0.81	1.11	0.55	0.39
0.50Hz	1.22	0.76	2.46	0.63



(a) (b)

Fig. 3. Trade-off between coherence function and measurement duration, pressure amplitude.

flow sensor using customized flow profiles generated by a pulmonary waveform generator (PWG-33, Piston Medical), which were sine waves with similar frequency and amplitude as tidal breathing. Then we calibrated the overall system against multiple frequencies from 5 Hz to 30 Hz by directing the airflow into a calibrated commercial airflow resistor (5cmH2O/L/s, Hans Rudolph Inc). Based on the results, the calibration coefficient of the device was computed as the ratio of the measured resistance value over the true resistance value.

IV. Results and Discussion

We conducted multiple tests to evaluate the performance of the device, which include characterization of the ultrasonic flow sensor, evaluation of the signal quality of varying oscillatory waveforms, assessment of the measurement accuracy using mechanical models, and assessment of the device repeatability with human subjects.

A. Characterization of the ultrasonic flow sensor

We tested the ultrasonic flow sensor against a series of standard sine waves generated by the PWG device, with a frequency between 0.25 Hz and 0.5 Hz and an amplitude from 0.5L/s to 2.0L/s to approximate real tidal breathing frequency and amplitude. Each sine wave contained 6 cycles and we computed the measurement error of peak expiratory flow (PEF) and forced vital capacity (FVC) for each cycle. The maximum error of PEF and FVC over 6 cycles is shown in Table 1 and Table 2.

The ultrasonic sensor is able to measure transit-time in a microsecond scale. From the results we observe that the ultrasonic sensor has high accuracy, and shows good linearity within the measurement range of forced oscillation tests.

B. Evaluation of the signal quality

We used the coherence function between pressure and flow sensor output to evaluate the signal quality of the device. The coherence function is defined as (3) [14]:

$$\gamma^2(f) = \frac{|G_{PV}(f)|^2}{G_{PP}(f) \cdot G^*_{VV}(f)}, 0 \leq \gamma^2 \leq 1, \qquad (3)$$

where G_{PV} is the cross-power spectrum between pressure and flow, and G_{PP}, G_{VV} are the auto-power spectrum of pressure and flow, respectively.

The coherence function reflects the linearity of the system and the quality of the output signal [14]. FOT guidelines generally require the coherence function to be $\geq$ 0.9 or 0.95 to ensure reliable impedance measurements [7][8]. Thus, we studied the impact of measurement duration and pressure amplitude on coherence function to provide guidelines to oscillatory signal design.

First we collected tidal breathing curves from 6 healthy subjects(IRB No. 825430-1) by asking them to wear a nose clip, seal their mouth around the mouthpiece and breathe normally into the prototype for 30 sec without the presence of excitation signal. Then we connected a mechanical model with calibrated resistive load of 2.5cmH2O/L/s(Hans Rudolph Inc) to the device and generated pressure oscillations of varying frequencies, amplitude, and duration. Following the approaches similar to those adopted by other studies [13][17], the excitation pressure and flow signals were recorded and combined with the breathing noise to get pressure and flow channel outputs, which were then used to compute the coherence function. Specifically, we first fixed the pressure amplitude to 0.2 kPa and studied the influence of measurement duration on coherence function. We then fixed the measurement duration to 5 sec and studied the impact of pressure amplitude. The results averaged over 6 subjects are shown in Fig. 3 .

From the results, we observe that the device is able to meet the recommended coherence threshold for all frequencies within a short measurement duration. Thus, it is able to provide quick and reliable FOT tests. Besides, there is a trade-off between measurement duration, pressure amplitude and coherence function. Increasing testing time or pressure amplitude could both increase the coherence function. The results indicate that we could compensate for the low SNR of the portable device with specially designed oscillatory waveforms to ensure reliable measurement result. Thus, the device can be programmed to generate oscillatory pressure waves sweeping over several frequencies, or multifrequency sine waves with tuned amplitude for each component to achieve reliable impedance measurement at multiple frequencies within a short duration of 30-40 sec.

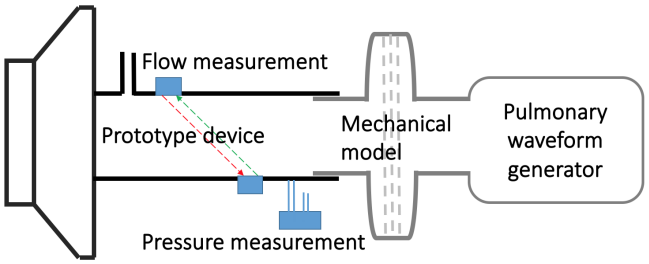


Fig. 4. Schematic representation of the experiment setup to test the prototype with mechanical models.

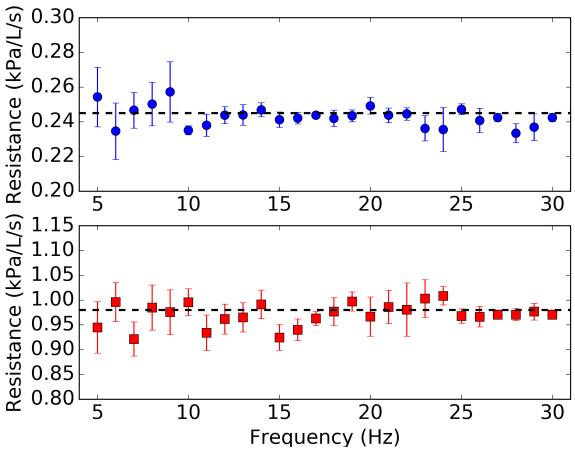


Fig. 5. Mean value and standard deviation of the measured resistance of the 2 mechanical models between 5-30 Hz. Scatter: measured value. Dash line: true value.

C. Assessment of measurement accuracy with mechanical models

We tested the accuracy of the device by measuring two mesh screen type mechanical models with calibrated resistive load of 2.5cmH2O/L/s, 10.0cmH2O/L/s (Hans Rudolph Inc), that approximated the resistance value of healthy adults and COPD patients. To further simulate real forced oscillation tests, we used the PWG to regenerate the breathing noise collected from 6 subjects while measuring the resistance of the mechanical models under an oscillatory pressure signal that was designed based on the signal quality analysis in section B. The pressure signal consisted of multiple single frequency sine waves sweeping from 5 Hz to 30 Hz. The duration of 5 Hz was set to 10 sec, and that of the other frequencies was set to 5 sec. The pressure amplitude was adjusted between 0.1-0.3kPa with an increased amplitude at 5 Hz to guarantee $\gamma^2 > 0.9$. Fig. 4 shows the schematic representation of the experiment setup.

The mean resistance and standard deviation over 6 measurements were computed for each frequency and Fig. 5 shows the result. Since the mesh-screen type models have pure resistant components, the estimated reactance value are of order 10^{-4} to 10^{-3} kPa/L/s for all frequencies, which is consistent with the true value.

From the results we observe that the maximum relative error of resistance measurement is below 10%, and the standard deviation of multiple tests with varying breathing noise pattern is small. The measurement accuracy satisfies the FOT guidelines [8].

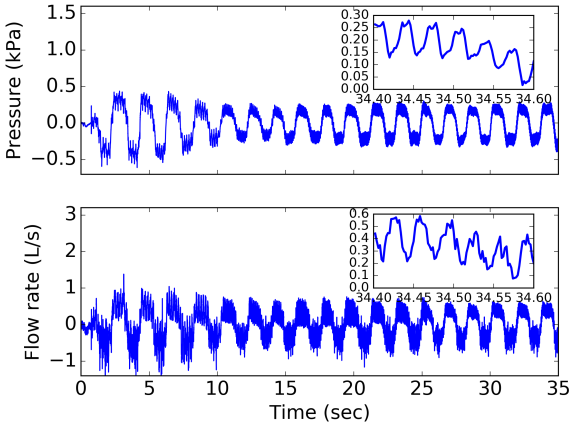


Fig. 6. Representative pressure and flow sensor output of human subject.

D. Assessment of measurement repeatability with human subjects

To assess the repeatability of the device, we measured the respiratory impedance of 6 healthy subjects (IRB No. 825430-1) and repeated the test 3 times for each subject during one visit.

During the tests, the subjects were required to wear a nose clip and seal their mouth around the mouthpiece. They were also asked to breathe normally into the device through a bacterial filter and support their cheeks to prevent vibration caused by the pressure oscillations. The duration of each test was 35 sec, and the excitation signal consisted of 5Hz, 10Hz, 15Hz, 20Hz, 25Hz, 30Hz single frequency sine wave with 10 sec for 5Hz and 5 sec for the others. The signal was designed to meet the coherence value threshold and to include 2-5 breathing cycles for each frequency. We would not allow the results and would redo the test if artifacts such as coughing, vocalization, swallowing occurred and the coherence function was lower than 0.9. Fig. 6 shows the representative outputs from pressure and flow sensor and Fig. 7 shows the separated high-frequency excitation signal and low-frequency breathing noise after filtering.

We then estimated the respiratory resistance and reactance value using (1), and corrected for the additional impedance introduced by the bacterial filter and mouthpiece. Fig. 8 shows the mean resistance and reactance value of the 6 subjects, as well as the standard deviation over 3 tests.

Through the tests, we find that the coherence value is highly dependent on the breathing pattern of each subject. It's generally more difficult to meet the 0.9 threshold at lower frequencies due to breathing interference, thus, the pressure amplitude at 5Hz needs to be increased to ensure

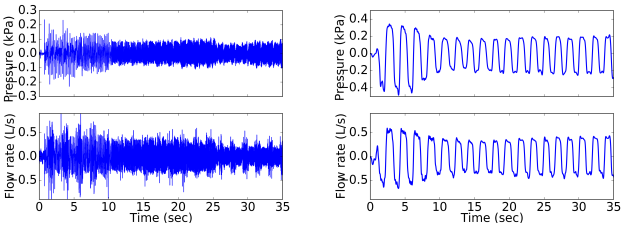


Fig. 7. Filtered signals using 3rd order Butterworth filter. (a) high frequency oscillation signal; (b) low frequency breathing noise.

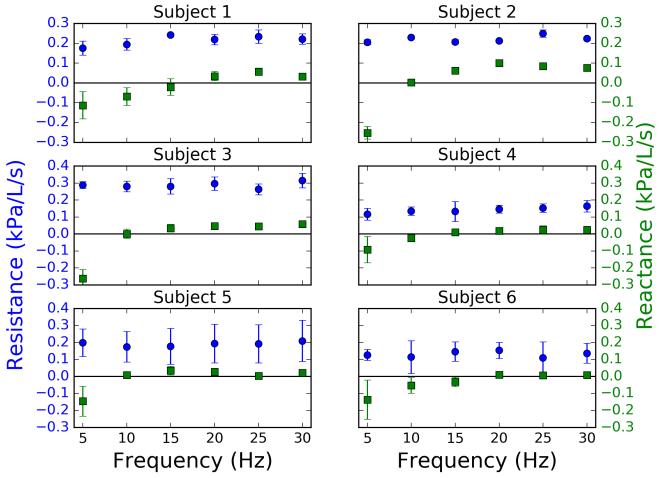


Fig. 8. Mean resistance and reactance value and standard deviation of 6 human subjects. Circle: resistance value. Square: reactance value.

sufficient SNR. Artifacts such as coughing, glottis closure, and strong turbulence created by forceful breathing will also result in the test failure.

From the result we observe that the resistance values are within the range of 0.1-0.4 kPa/L/s and show a frequency-independent pattern. The reactance values are negative at 5 Hz and gradually increase to some positive values with the zero-crossings between 5-18Hz. The measurement result is consistent with clinical empirical values of healthy subjects. The typical standard deviation of resistance and reactance value is lower than 0.08 kPs/L/s and 0.005 kPa/L/s, respectively. The results from multiple tests verified the good repeatability of the device.

V. Conclusion

In this paper, we verified the feasibility of a portable forced oscillation device for pulmonary function tests. We developed and optimized the signal processing algorithm for impedance estimation, and studied the impact of frequency, measurement duration, pressure amplitude on signal quality. The device was tested against both calibrated mechanical models and healthy human subjects. It shows good repeatability during multiple tests, and is able to achieve a coherence value > 0.9 and a relative measurement error $< 10\%$ for all frequencies between 5-30Hz. The result shows that the prototype meets the guidelines for FOT tests, and satisfies the clinical requirements for point-of-care pulmonary function testing. In the future, we plan to carry out a large-scale clinical trial in India with Asthma and COPD patients and further demonstrate the feasibility of the device.

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