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A System for Recording High Fidelity Cough Sound and Airflow Characteristics

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Abstract—Cough is considered an early sign of many respiratory diseases. Recently, there has been increased interest in measuring, analyzing, and characterizing the acoustical properties of a cough. In most cases the main focus of those studies was to distinguish between involuntary coughs and ambient sounds over a specified time period. The objective of this study was to develop a system to measure high fidelity voluntary cough sounds to detect lung diseases. To further augment the analysis capability of the system, a non-invasive flow measurement was also incorporated into the design. One of the main design considerations was to increase the fidelity of the recorded sound characteristics by increasing the signal to noise ratio of cough sounds and to minimize acoustical reflections from the environment. To accomplish this goal, a system was designed with a mouthpiece connected to a cylindrical tube. A microphone was attached near the mouthpiece so that its diaphragm was tangent to the inner surface of the cylinder. A pneumotach at the end of the tube measured the airflow generated by the cough. The system was terminated with an exponential horn to minimize sound reflections. Custom software was developed to read, process, display, record, and analyze cough sound and airflow characteristics. The system was optimized by comparing acoustical reflections and total signal to background noise ratios across different designs. Cough measurements were also collected from volunteer subjects to assess the viability of the system. Results indicate that analysis of cough characteristics has the potential to detect lung disease.

Keywords—Cough recording system, Cough sounds, Cough airflow, Lung sounds.

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INTRODUCTION

Diseases of the respiratory system are considered to be an important cause of illness and death in man. Respiratory disease ranges from mild such as the common cold to life-threatening such as pulmonary emphysema. Cough is considered one of the most common symptoms of respiratory disease.[7] It has several practical functions which are usually occasional and causal in nature and are normally associated with inflammation of the respiratory system or aspiration of ingested materials. Cough can be initiated either voluntary or involuntarily. There are various physiological parameters associated with cough which can be measured including electromyography (EMG) of the respiratory muscles, pulmonary pressures, lung volume changes, airflow patterns, and cough sound intensity.[5] It has been shown that healthcare professionals are able to recognize some of the qualities of cough sounds, but are not proficient at making accurate diagnoses based on this information.[13]

The usefulness of applying signal processing techniques to analyze cough sounds of normal subjects and subjects with lung disease has been examined over the last two decades. Vrabec *et al.*[17] developed a non-invasive method for sound recording, sound analysis and the measurement of the time duration of coughs. They analyzed sounds not only generated during coughing but during crying and breathing based on a database that had been recorded on magnetic tape. In their study, the analog sound recordings were digitized with an A/D converter and their frequency content was determined based on the fast Fourier transformation (FFT) of the data. In the same year, one of the earliest designs of a portable cough sound recorder was described by Kohler *et al.*[8] These investigators developed a cough recorder linked to a portable commercially available actigraph to record both the acoustic characteristics of a cough and the movement of the ventral thorax. An analysis based on signal processing algorithms was used to differentiate cough sounds from background noise for 10 volunteer subjects. One year later, Murata *et al.*[11] used an electret microphone to record cough sounds which were stored on a digital audiotape recorder having a bandwidth between 20 Hz and 20 kHz. The cough sounds were generated by patients with either productive or non-productive cough resulting from chronic airway diseases. The sounds were recorded under free field conditions and were compared with voluntary coughs from healthy subjects using spectrograms and time-expanded waveform analyses.

The work presented by Van Hirtum and Berckmans[16] applied an artificial intelligence approach for classifying different kinds of human coughs. They performed free-field acoustical measurements with a standard multi-media microphone on two groups of subjects to determine if there were differences between spontaneous and voluntary coughs. Experimental error rates of their classifier were established using different neural and fuzzy classification networks. In another study by Korpas *et al.*,[9] voluntary cough sounds were recorded using a microphone and saved on analog tape before they were digitally processed. These investigators studied adults and children with bronchial asthma, and the frequency content of the recorded cough sounds was examined using FFT with an overlapping technique.

Recently, other investigators have worked to develop more specialized systems for automatic cough monitoring that could be commercialized. A system for quantifying cough frequency content was evaluated by Coyle *et al.*[2] In this system, a microphone placed over the throat was integrated with a commercially available ambulatory physiologic monitoring system (*Vivometrics Lifeshirt*™) to measure respiratory patterns. An automated algorithm for cough recognition was developed, and the complete system is now commercially available for cough monitoring.[14] The Hull automatic cough counter (HACC) is another system that is capable of recording ambulatory sound over 24 h using a free field microphone.[1] The HACC has adopted a classification algorithm for human coughs using an adaptive neural network. In an additional study, Matos *et al.*[10] used hidden Markov models as the basis of a cough classification algorithm. This system has been implemented and described as the Leicester cough monitor (LCM). The LCM is capable of recording and analyzing cough sounds, either continuous or ambulatory for 24 h using audio recordings made with a miniature microphone attached to a digital audio recorder.

A more complete description of new developments in the objective assessment of cough have been reviewed and compared by Smith and Woodcock.[14] The majority of the techniques mentioned above, in addition to others, are described along with methods and equipment to record cough sounds. Most methods used free field sound measurements to determine whether the recorded sound wave was either a cough or a different acoustical event. The only system that has been designed to measure both sound and flow during coughing in a tube was the system designed earlier by Thorpe *et al.*[15] However, the system was not mobile, and standardization and sound reflections testing procedures were not considered.

In the past, there have been some algorithms used to classify coughs as either normal or abnormal, while others only identified and counted the number of coughs. Coughs recorded in a free field, however, may not provide standardization of the measurement due to several factors. For instance, variations in geometry of the mouth and lips during repeated coughs can alter the characteristics of the cough signal. Reflections of the cough sounds returning from nearby structures, unwanted sounds and noise generated in the external environment also affect the fidelity of the recorded cough sound. These factors can alter the accuracy of the measurement and confuse the classification process.

The objective of this study was to develop and test a high fidelity system to record voluntary cough sound and airflow characteristics. These characteristics can be used for pulmonary function testing and screening to determine if a subject has pulmonary disease.[6] Moreover, the same system could also be used to identify individuals by their cough characteristics.[4]

METHODS

System Description

A system was developed to measure the sound pressure waves and airflow that propagate through the mouth during a cough. The schematic diagram shown in Fig. 1 describes the main components of the cough measurement system. In this system, a cylindrical mouthpiece was attached to a 2.54 cm diameter metal tube. A 6.35 mm microphone (Model 4136, Bruel & Kjaer, Denmark) was mounted at a 90° angle with its diaphragm tangent to the inside of the metal tube. A 2.54 cm diameter flexible tube was attached to the metal tube on the end opposite to the mouthpiece. A pneumotach (Model Fleisch No. 2, Fleisch, Epalinges, Switzerland) and differential pressure transducer (Model 239, Setra Systems, Boxborough, MA) were

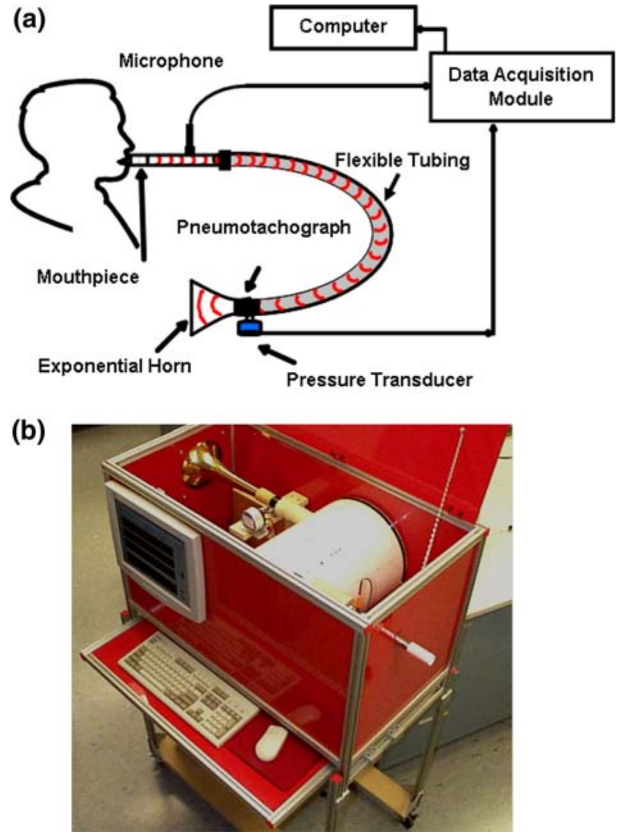

FIGURE 1. The high fidelity system used to simultaneously record sound pressure waves and airflow during a voluntary cough. (a) A schematic diagram of the system and (b) a system photograph.

employed at the terminal end of the flexible tube to measure airflow during a cough. Airflow was determined by measuring the pressure drop across a capillary type pneumotach with a known resistance. Flow was measured at the distal end of the flexible tube (Fig. 1) so that any acoustical reflections from the capillary tubes in the pneumotach would be minimized due to losses in the transmission path back towards the microphone. The system was terminated with an exponential horn (trombone bell) to reduce acoustic reflections. A custom cart was constructed to house the recording system.

A software "virtual instrument" was designed using LabVIEW 7.0 (National Instruments, Austin, TX) to capture the airflow and acoustical properties of voluntary coughs. The instrument was flexible and enabled the user to control many of the acquisition parameters. These parameters included the sampling frequency, sampling time, analog high-pass and anti-aliasing filters, input range, and triggering considerations.

Cough signals were typically digitized at a sampling rate of 65,536 Hz using a 16-bit data acquisition board (Model PCI 4451, National Instruments, Austin, TX). The signals were recorded for one second after a cough was initiated. A pre-trigger period of 0.05 s was included to ensure that the initial portion of the cough response was recorded. The DC offset for the pressure transducer signal was acquired by the software prior to each measurement. The flow waveform was DC coupled and the DC offset was subtracted from each measurement. The sound signal was AC coupled in order to remove the low frequency pressure voltage from the signal which, otherwise, would have clipped the readings. The AC coupling capacitor was embedded on the data acquisition board and acted as a high-pass filter with a -3 dB cutoff frequency of 3.4 Hz, and a -0.01 dB cutoff frequency at 70.5 Hz. Analog anti-aliasing filters, embedded on the board, provided a sharp roll-off and very little phase error with a rejection of over 85 dB above 35 kHz. These filters were used to assure proper acquisition of the acoustic signal and to remove noise outside the microphone range. The frequency response of the Bruel & Kjaer condenser microphone ranged between 20 Hz and 35 kHz ($\pm$1 dB). The recorded signals enabled accurate spectral analysis in the frequency domain between 50 Hz and 25 kHz.

Following their acquisition, the acoustic and airflow characteristics of a cough were displayed vs. time along with the flow–volume relationship. The dataset was saved using the custom LabVIEW program. The data was then post processed using Matlab 6.1 (The MathWorks, Inc., Natick, MA).

System Optimization

Various experimental tests were performed to optimize the system design. In order to assess the performance of various designs, a sound horn (Model PD-30, Phoenix, AZ) was coupled to the mouthpiece of the recording system to generate consistent test signals. In the first set of tests, a five-volt peak-to-peak pulse (133.15 dB peak SPL) was generated and amplified with a power amplifier (Model DA-M10, Mitsubishi, Japan), then used as an input to the sound horn. All sound pressure level (SPL) measurements expressed in dB are referenced to 20 μPa unless otherwise noted. The incident acoustic pulse and subsequent unwanted reflections from the end of the tube were recorded. The autocorrelation of the recorded signals, including the initial pulse and subsequent reflections, was evaluated. The temporal delays of the reflections appeared as peaks in the autocorrelation. The peaks were separated by the delay times. The total energy within the initial pulse and the reflections was calculated and compared to determine the relative amplitude of the reflected signal. The effects of a range of parameters on the magnitude of the reflections were examined. Those parameters included several types of flexible tube materials, various tube lengths, and different types of anechoic terminations. Although various types of tubing materials were tested, only the results of the best four, Latex (L), Norprene (N), Braided Silicone (BS) and Santoprene (S), were reported. All tubing types had an internal diameter of 2.54 cm. Each tube type also had a wall thickness of 6.35 mm, except the BS tubing which had a wall thickness of 9.14 mm. Tubing length was also varied to determine its effect on system performance. In addition, anechoic terminations including an exponential horn and a conical wedge were tested to evaluate their effect on acoustical reflections. Wedge position was varied to determine the placement at which minimum reflections occurred. The optimal system (OS) was defined as having the configuration that provided the least reflections.

In another set of tests, the total signal to background noise ratio (TBR) was measured, and the OS system was evaluated and compared with that of a conventional free field recording method (CRM). The free field measurements were made with the same microphone used in the OS system located at the same distance and orientation with respect to the sound source. First, background noise was recorded for the OS and the CRM. Then, white noise (120.53 dB SPL with a flat frequency spectrum from 25 Hz to 25 kHz) was applied to the sound horn for a full second, and sound recordings were made under both arrangements. Measurements of total signal were a combination of both the white noise and background noise energy. The resulting SPLs were measured for each configuration and the TBRs were calculated.

System Calibration

Due to the distance between the mouthpiece and pneumotach, it was assumed that the flexible tube would delay and filter the flow signal (Fig. 1). In order to describe the characteristics of the delay and filter, a second pneumotach was temporarily introduced at the mouthpiece for testing purposes. Initially, both pneumotachs were calibrated by delivering 35 constant flow waveforms, each having a volume of 12 L. The flows ranged from 0.8 to 14 L s^{-1} and were generated using the piston pump described by Reynolds *et al.*[12] For each waveform, a calibration factor multiplied by the integral of the pneumotach signal was set equal to 12 L. A curve fit was applied to these data yielding an expression for airflow as a function of the pressure across each pneumotach. Flow measured at the mouthpiece was considered to be the ''true'' flow. The flow measuring system was modeled with a delay followed by a low order autoregressive moving average (ARMA) filter. The parameters of the model were determined by exciting the system with a noise signal generated by a voltage controlled pump. The delay and filter order were determined using standard ARMA modeling techniques (System Identification Toolbox, MATLAB, Mathworks, Inc.). The output of the second pneumotach was shifted by the delay and a second order digital filter was used to describe the transfer function between the proximal pneumotach and the delay compensated distal pneumotach. Before the proximal pneumotach was removed, the calculated delay and filter were tested using several coughs spanning a range of flow amplitudes and frequencies. The linearity of the system was demonstrated by the fact that the calculated delay and filter, when applied to these test cough flow recordings measured at the proximal pneumotach, accurately predicted the flow measured by the distal pneumotach. After testing, the proximal pneumotach was removed and the inverse of the calculated filter was applied to the recorded airflow measurements to estimate the actual airflow at the mouth. This provided an estimate of ''true'' flow from flow recorded at the distal end of the tube.

The microphone was calibrated using two pistonphones (Models 4228 and 4230, Bruel & Kjaer, Denmark) which produced 124 dB SPL at 250 Hz, and 94 dB SPL at 1 kHz, respectively.

Cough Measurement Protocol

Figure 2 shows the graphical user interface (GUI) of the system used to record the acoustical and airflow characteristics of a cough. Four waveforms were displayed following each cough including the sound pressure, airflow, air volume, and the flow-volume curves. In the final evaluation of the OS, voluntary control coughs were recorded at the Pulmonary Clinic of West Virginia University, Ruby Memorial Hospital. The protocol was reviewed and approved by the local institutional review board, and all participants gave written informed consent. Spirometery measurements were used to confirm that each test subject was from the control population. To ensure that each person had the same lung volume history, prior to a cough, they were asked to inhale to total lung capacity (TLC), exhale passively to functional residual capacity (FRC), inhale a second time to TLC and hold their breath for 3 s. Then form a seal with their teeth and lips around the mouthpiece and cough.

Repeatability of Recorded Voluntary Coughs

Volunteer subjects ($n = 4$) were asked to perform a voluntary cough 3 times on 7 different occasions. The age of the volunteers ranged from 28 to 65 years. In order to demonstrate the repeatability of cough sound

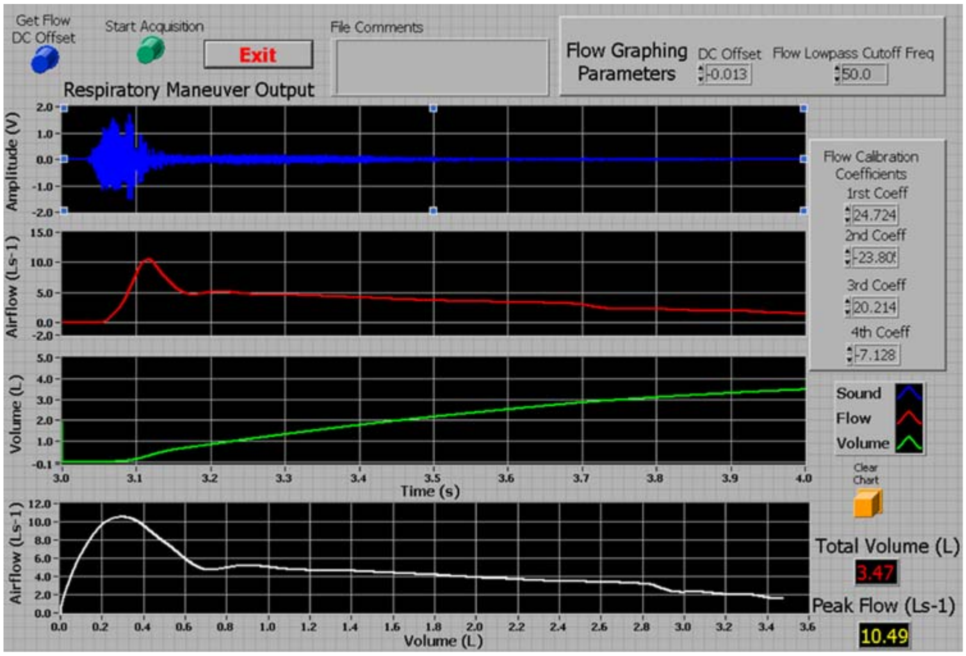

FIGURE 2. Graphical user interface (GUI) used to display and record cough flow, volume, and acoustical signals.

and airflow characteristics measured with the new system, a total of 21 coughs were recorded for each test subject. Five different sound and airflow features were extracted for each cough: (1) peak airflow during a cough (F_P); (2) average airflow during a cough (F_{avg}); (3) total cough volume (V_t); (4) value of β which produces the best fit of the relationship Amplitude $= K/f^{\beta}$ with the spectrum of the cough sound pressure wave (β); and (5) the ratio of the cough airflow length to cough sound length (L-Ratio). The mean and standard deviation of each feature were calculated for the coughs from each subject.

RESULTS

Generally, when recording cough sounds, it is preferable to have a high signal to noise ratio and minimal interference from sound reflections. An initial study was performed to determine which material was able to provide the lowest acoustic reflection amplitude or greatest attenuation of the cough sound after it traveled beyond the microphone. For this experiment 3.0 m lengths of several types of flexible tube material were used. Table 1 shows a comparison of the reflection amplitudes of a simulated impulsive sound for L, N, BS and S type tubing and the arrival time of the first reflection. Results showed that the latex tube performed optimally in those tests. Though all four tubes had similar reflection times (L = 19 ms, N = 19 ms, BS = 18 ms, S = 19 ms), the relative reflection

TABLE 1. Arrival time of the first reflection and its relative amplitude to the incident wave are shown after a sound pulse was delivered through the recording system with various material types of flexible tube.

Material type	Arrival time (ms)	Reflection amplitude (%)
Latex	19	6.6
Norprene	19	15.0
Braided silicone	18	38.0
Santoprene	19	9.3

Note: Flexible tube length was 3.0 m and no anechoic terminations were used during this round of tests.

amplitude was much lower with the latex tube (6.6%) when compared with the other materials. Figure 3 compares the normalized autocorrelation function of the test sound pressure waves recorded with the least reflective (L) and most reflective (BS) tubing. The initial peak at time 0.0 represents the initial pulse while the secondary peaks at latter times represent the subsequent reflections. The secondary peak (denoted with an ○) occurs at the arrival time of the reflected wave at the microphone. Based on the experimental evidence, the latex flexible tube material was selected for use in the cough recording system.

Additional experiments were performed to determine the optimum length of the latex tube based on the reflection amplitude and delay time. Longer tubing length produced less reflection but also presented larger resistance to airflow. The elevated resistance experienced during trial coughs caused subject discomfort. For this reason, lengths of more than 4.0 m

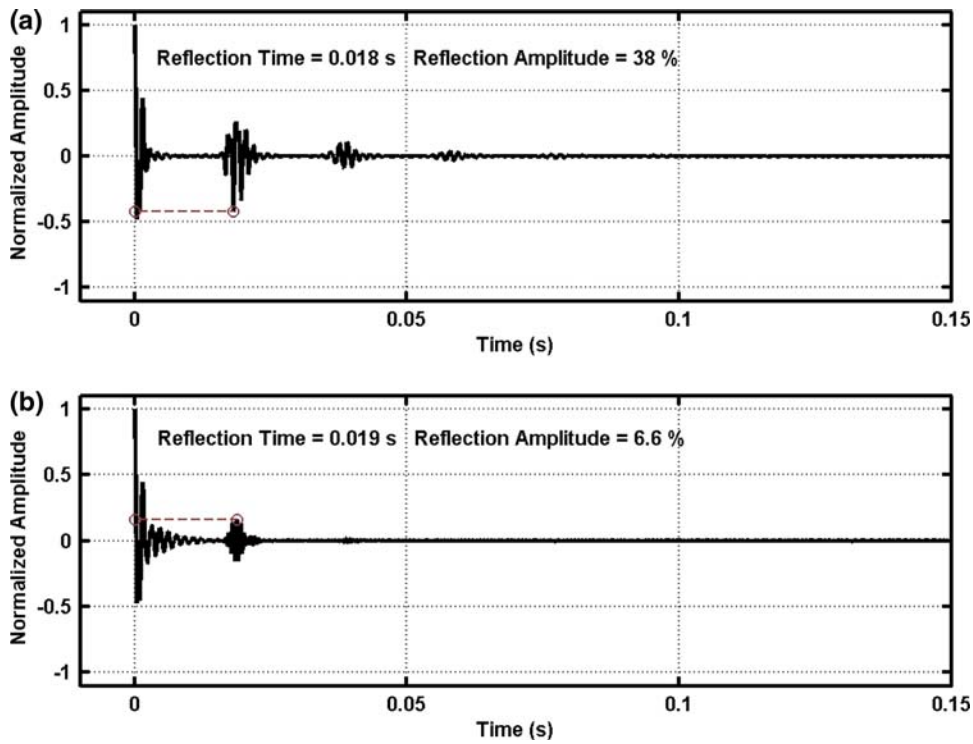

FIGURE 3. The normalized autocorrelation function of sound pressure wave recordings for two different materials: (a) Braided silicone tube, and (b) latex tube. Primary peak at time 0 represents initial pulse, secondary peaks represent reflections. Highest amplitude of secondary peak (denoted with an ○) indicates reflection time-lag.

were disqualified from consideration as a component of the system. The 4.0 m length provided an optimal balance between reduced reflections and subject comfort. When using this length of tubing, the reflection amplitude was further reduced to 1.5%.

A final experiment was performed to determine if an acoustic termination added to the distal end of the flexible tube would improve the system performance. It was found that the addition of an exponential horn decreased the amplitude of the reflections observed with the 4.0 m latex tube from 1.5 to 0.65%. The conical wedge had little to no effect on the amplitude of the reflection at any position inside the exponential horn.

Test results indicated that a 4.0 m latex tube terminated with the exponential horn provided the OS. Using this configuration, this system exhibited a reflection time lag of 27 ms and reflection amplitude that was equal to 0.65% of the original test signal.

The tests comparing the total to background ratios of the OS with a CRM indicated the higher signal strength in the OS measurements markedly increased the fidelity of the recording. The TBR for the OS was 49.7 dB while the ratio was 21.0 dB for the CRM.

Figure 4 shows typical curves for the sound and airflow waveforms recorded during a voluntary cough of a control subject. The cough airflow curve (Fig. 4a), initiated during the expulsion phase with a rapid flow, increased until reaching its peak at 9.5 L s^{-1}. The rise time of the airflow relationship was approximately 0.05 s, then, about 0.375 s after the peak, the airflow dropped to 10% of its peak value. The sound wave shown in Fig. 4b was produced during the same cough. It also reached its highest intensity during the expulsive phase and continued over a period of about 0.3 s.

The cough sound measurements of the four volunteer control subjects were similar in terms of frequency content. The energy within the coughs decayed nearly exponentially as the frequency content approached 25 kHz. An example of power spectral density curves for two control coughs and a background measurement is shown in Fig. 5.

A spectrogram of the cough sound of a control subject was computed to provide information about the cough sound in the time-frequency domain. Figure 6 shows a typical plot for a control subject where the red and orange colors represent higher amplitudes. The spectrogram contained a wide range of frequency components in the 50 Hz to 25 kHz range. The amplitudes were higher at lower frequencies (50–100 Hz) and existed for a longer time period than they did in the higher frequency ranges.

It has been demonstrated that the flow-volume relationship achieved during a voluntary cough represents a partial maximum expiratory flow–volume maneuver that is effort independent. It follows, therefore, that the airflow during a cough should be repeatable when the cough is initiated from the same lung volume.[3] The repeatability of recorded voluntary coughs using the system described in this study is illustrated in Table 2. The mean and standard deviation of main features characterizing the cough sound and airflow signals for the subjects within the test set are shown. It can be seen that the average value of each feature varied between subjects (large inter-class variation), but the variation in feature values for a given subject was, in most cases,

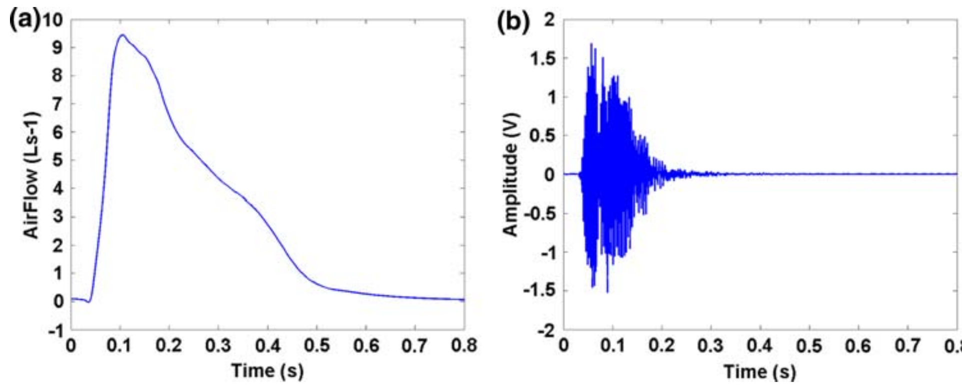

FIGURE 4. Airflow (a) and sound pressure wave (b) measured during a voluntary cough for a control subject.

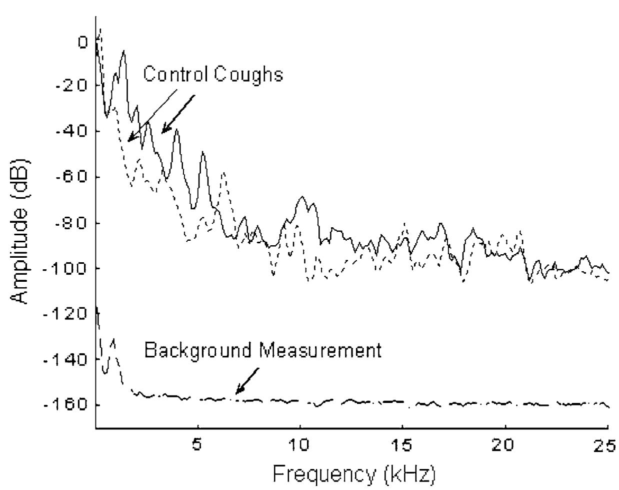

FIGURE 5. Power spectral density curves for control coughs from two subjects and a background measurement. For comparison, the reference value for the dB measurements was chosen as the peak amplitude of the frequency spectrum representing the control cough designated by the solid black line.

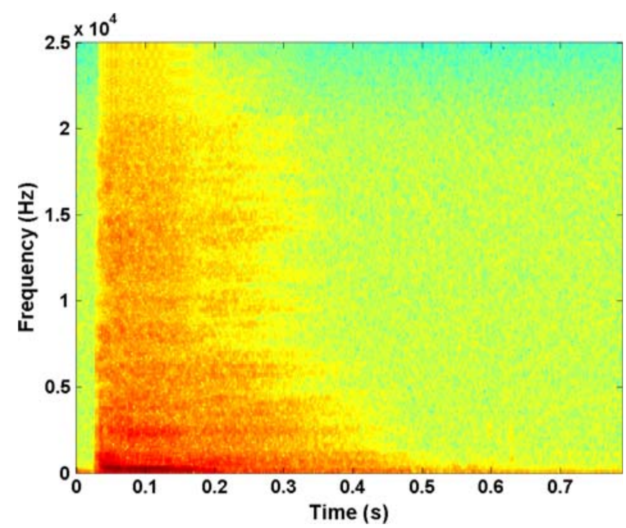

FIGURE 6. Spectrogram showing the time–frequency domain relationship of a cough from a control subject.

TABLE 2. The mean and standard deviation (M ± SD) of five different parameters extracted from the cough sound and airflow characteristics of 21 coughs from each of four subjects.

Subjects	F_p (L s^{-1}) (M ± SD)	F_{avg} (L s^{-1}) (M ± SD)	V_t (L) (M ± SD)	β (M ± SD)	L-Ratio (M ± SD)
1	10.68 ± 0.4049	3.724 ± 0.0669	3.649 ± 0.0671	2.5161 ± 0.1729	1.0075 ± 0.0399
2	12.52 ± 0.4789	3.322 ± 0.2007	2.777 ± 0.1299	2.5767 ± 0.1115	0.80092 ± 0.0584
3	11.72 ± 0.3932	3.950 ± 0.0749	3.8618 ± 0.0746	2.012 ± 0.1301	1.0341 ± 0.0310
4	9.90 ± 0.3346	4.337 ± 0.176	4.2137 ± 0.1758	2.522 ± 0.0914	0.94392 ± 0.0407

Note: The parameters include: (1) peak airflow during a cough (F_P); (2) average airflow during a cough (F_{avg}); (3) total cough volume (V_t); (4) value of β which produces the best fit of the relationship amplitude $= K/f^{\beta}$ with the spectrum of the cough sound pressure wave (β); and (5) the ratio of the cough airflow length to cough sound length (L-Ratio).

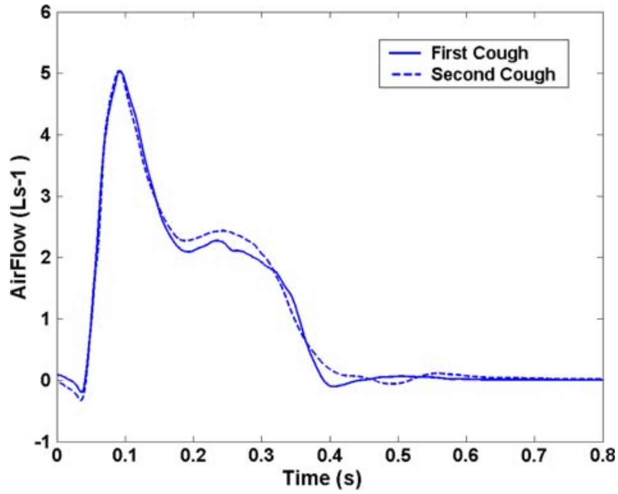

FIGURE 7. Airflow measurements of two successive coughs for the same control subject. The two airflow signals are very similar which illustrates the repeatability of airflow measurements during voluntary coughs.

quite small (small intra-class variation). The coefficients of variation of each feature determined from the coughs of four test subjects ranged between 3.37 and 7.20%. This demonstrates the repeatability of coughs produced by a given test subject and the consistency of the recording system. Due to this high degree of repeatability, algorithms were developed that could identify individuals from their voluntary coughs.[4]

To illustrate the similarities of the individual waveforms, the airflow signals measured during two voluntary coughs of one of the subjects using the cough recording system is shown in Fig. 7. In addition, the spectrograms of the cough sounds generated during the coughs shown in Fig. 7 are shown in Fig. 8. Many similarities in the time-frequency domain for cough sounds from the same subject were observed. By contrasting Fig. 7 with Fig. 4a and Fig. 6 with Fig. 8, the inter-class variation can be observed graphically for cough airflow and sound, respectively.

CONCLUSION

A high fidelity mobile system was designed and tested for measuring the characteristics of a voluntary cough. Minimized reflections and an improved TBR were achieved by choosing an optimal length and type of flexible tubing along with terminating the tube with an exponential horn.

Several experimental tests were performed using different tube materials, lengths, and terminations in order to reach the final design. The configuration expressed in the System Description was optimized by using a latex flexible tube (4.0 m length, 6.35 mm wall

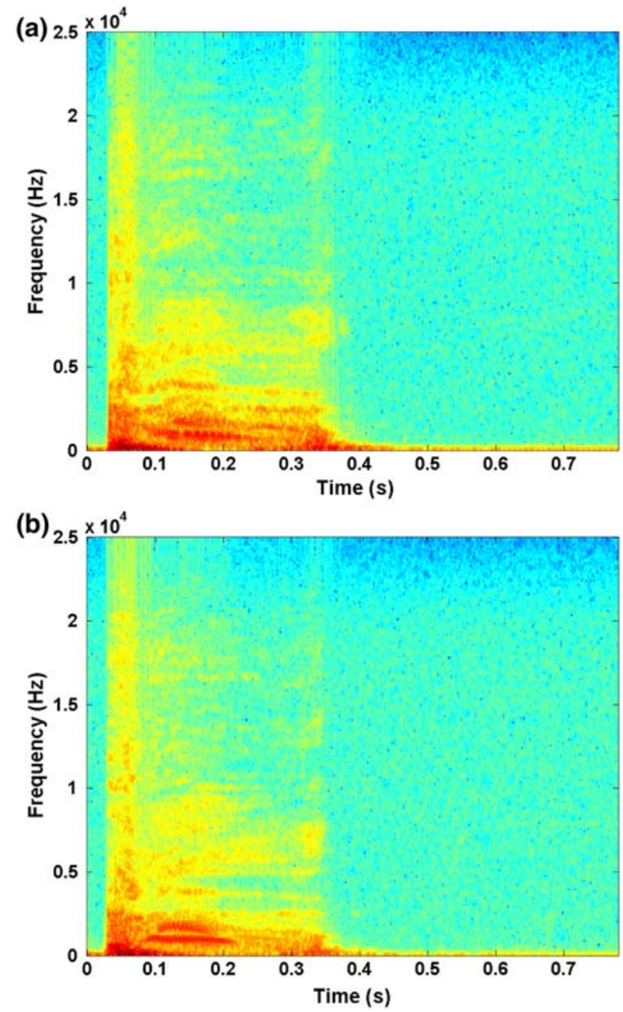

FIGURE 8. Spectrograms of cough sounds generated by the same control subject.

thickness) terminated with a trombone bell. The system was further evaluated with control subjects performing voluntary cough measurements using a well-defined recording protocol at a local pulmonary clinic. The sound and flow parameters of coughs from volunteer subjects were repeatable when the measurements were made with the optimally designed system under control conditions. The repeatability in voluntary coughs and their measurements was demonstrated by the modest variation in the main features characterizing the coughs of individual test subjects. This was true even though there were obvious differences in the average values of features between test subjects.

In summary, there are several advantages of the recording system described in this study when compared to systems using a microphone at a set distance from a test subject: (1) the signal to noise ratio of the new system is improved; (2) the sound pressure levels measured for each cough are higher since the sound is primarily propagated in only one direction; (3) the effects of acoustical reflections from objects in the immediate vicinity are reduced; (4) the position of the microphone with respect to the mouth of the test subject is precisely located for coughs recorded on different occasions; (5) the position of the mouths and lips are the same for all coughs; and (6) a combination of both sound and airflow measurements can be made for individual coughs.

In the future high fidelity measurements of acoustical and airflow characteristics of a voluntary cough combined with signal analysis techniques could be used to determine if a cough was produced by a normal subject or a subject with lung disease.

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