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Automatic Wheeze Detection System as Symptoms of Asthma Using Spectral Power Analysis

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Abstract

People all over the world live and die from, respiratory disorders. It doesn't matter who suffer may be young or old. This respiratory problem they have from birth or develop in certain environmental condition over the years. Wheezes are more musical respiratory sounds as compared to normal respiratory sound, which usually due to pathological condition of the respiratory system especially irregularities in pulmonary obstruction, such as asthma and chronic obstructive pulmonary disease (COPD). Identification of the Wheeze sound is the first step in controlling the asthma. Using that easy to understand the disease progress, and how close research to developing new test and treatments and what still needs to be done. Although many studies have addressed the problem of wheeze detection, a limited number of scientific works has focused on real time detection of wheeze sound. As day to day increasing number of asthmatic patient there is a need of automatic monitoring of the wheeze sound to assist the physicians in diagnosing and monitoring the patient. The purpose of this study was to develop automatic wheeze detection systems with spectral power estimation in real time to assist the physicians in diagnosing and monitoring the patient.

Keywords: Asthma; Respiratory sound (Rs); Spectral power estimation; Wheezes

Introduction

In Nineteen Century Laennec noticed that respiratory sound generated by the lungs during inspiration and expiration contains information about respiratory diseases who also invented the relationship between human respiratory diseases and respiratory auscultation. He then invented stethoscope in 1921 which enabled physicians to listen to respiratory sounds of their patients and detect any symptomatic signs [1-6]. Wheeze to occur as symptoms of respiratory diseases result of airway obstruction and flow limitation at critical flow rates in frequency range between 100 Hz to above 1000 Hz. It is present dominantly during expiration and it lasts from 80 – 250 ms [7]. Wheezing is not associated with asthma only, but to other pulmonary pathology such as chronic obstructive pulmonary disease (COPD), bronchiolitis [8-17]. For diagnosis the asthmatic patient physician normally used stethoscope as it is conventional method. But auscultation using stethoscope is subjective method and it is high possibility of false diagnosis because it need well trained physician to recognize abnormalities and ability to differentiate between the sound patterns [1-6]. As day to day increasing number of asthmatic patients there is a need of automatic monitoring of the wheeze to assist the physicians in diagnosing and monitoring the patient.

According to literature survey first computerized wheezes detection system which is based on time-expanded waveform. This system based on searching the number of peaks or by combining spectra with concerning the amplitude, duration, and pitch range of the wheeze [18]. Clinically several methods are used for detecting the wheeze such as spirometer and other pulmonary function test [19]. But these are very time consuming methods and patient need to put extra efforts. The purpose of this study was to develop an automatic system based on classical method of spectral analysis for analysing and visualizing asthmatic respiration sound which can monitor a health condition in real time using acoustical information and detect an abnormal symptom. In this paper we suggest the method that can easily identifies the wheeze sound without physician depending upon the frequency components presents in the particular diseases that help to treat the patient.

Materials and Method

Data acquisition

The real time respiratory sound recordings were carried out in a laboratory with the subjects in sitting position. Single electrets condenser microphone (ECM-77B, Sony, Inc., Japan) was inserted into a hemispherical rubber chamber of stethoscope, and placed over a trachea. We applied the sensor for collecting the respiratory data at trachea because the trachea to be a better location for analysing respiratory sound than the lung while claimed that trachea is reliable because all air-propagated lung sound from the two lungs integrates in trachea. Moreover, more frequency information is preserved at the trachea, as the chest wall filters out higher frequencies. The respiratory amplifier is ultrahigh-gain preamplifier, which is fed by electrets microphone is actually an extremely sensitive audio preamplifier that exploits the fact that moving air close to a microphone causes an overload in the audio output. The recording software Agilent Technologies 1.8.5.0 (Model No. U2352A, serial No. TW50471004) was used and the RS recordings were saved in a mono channel at sampling frequency (*Fs*) of 10000 Hz. was used.

Data processing

In data processing we first applied the band pass filter to remove the heart sound signal. Heartbeat is an unavoidable source of interference for respiratory sound recording that when it occurs, changes both frequency and time characteristics of the respiratory sounds. Since heart sound and respiratory sound fall in the same frequency region;

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it is difficult to suppress heart sound signal from respiratory sound signal, so it is difficult to analyse the frequency components presents in respiratory sound signal in abnormal and normal condition. Most of the researcher not takes care of heart sound signal as a noise in respiratory signal which affect the analysis and extraction of the respiratory sound. Then we plot the frequency response, power spectrum and spectrogram of the filtered signal [20].

Filtering

For removing the heart sounds in respiratory sound is obtained by using a Kaiser window FIR band-pass filter [50 Hz, 2500 Hz]. Nevertheless, a high-pass filter at 100 Hz is not a good solution in so far as the main components of respiratory sounds are also located in this frequency range. Kaiserord uses empirically derived formulas for estimating the orders of low pass filters, as well as differentiators and Hilbert transformers. Estimates for multiband filters (such as band pass filters) are derived from the low pass design formulas [21].

The design formulas that underlie the Kaiser window and its application to FIR filter design are as given under

$$\beta = \begin{cases} 0.1102(\alpha - 8.7), & \alpha > 50 \\ 0.5842(\alpha - 21)^{0.4} + 0.07886(\alpha - 21), & 50 \ge \alpha \ge 21 \\ 0, & \alpha < 21 \end{cases}$$

Where $\alpha=-20\log_{10}\delta$ is the stop band attenuation expressed in decibels (recall that $\delta_p=\delta_s$ is required).

The design formula is

$$n = \frac{\alpha - 7.95}{2.285(\Delta\omega)}$$

Where n is the filter order and $\Delta\omega$ is the width of the smallest transition region

Frequency response

The frequency response of a digital filter can be interpreted as the transfer function evaluated at $z=e^{j\omega}$. You can always write a rational transfer function in the following form.

$$\mathrm{H}(e^{j\omega}) = \frac{\sum_{k=0}^{M-1} b(k)e^{-jwk}}{\sum_{l=0}^{N-1} a(l)e^{-jwl}}$$

Spectral power

The spectral power is computed for each of the selected samples in the frequency range 50 to 2000 Hz. A straight forward method which mostly used is discrete Fourier transform (DFT) to calculate the power with high resolution. It does not require setting the further parameters but it is computationally expensive. The resulting spectral vectors are not constant for all respiratory signal samples and impractically high dimension (a few thousands). Frequency resolution as well as dimension depends upon the length of original respiration signal. The respiration duration vary depend on natural variation. Averaging techniques reduces the dimension of the spectral vector over a present number of spectral samples.

For calculating spectral power with less computational cost is possible by using the Welch method of spectral estimation [20]. In this method we have a choice to select the type of window, width of the window and amount of overlap of window. We select and applied Kaiser Window with 50% overlap and window width is chosen such that the frequency resolution of the Welch spectrum equal to the average db in width of the DFT spectrum.

The sound power depends on the flow of air during inhale and exhale. The flow respiration sound tends to vary due to irregularities in pathological condition of the respiratory system, such as variations in respiration detect using depth and respiration frequency.

The variance of the power spectral density estimate ($\overset{w}{P}$) without segmentation at frequency f_b is equal to the square of its contents [20].

$$\mathrm{var}\{\overset{w}{P}(f_b)\} = P^2(f_b)$$

Where $P\ (f_b)$ is the expectation value of $\overset{w}{P}(f_b)$. In case of 50% overlap between successive data segments considerable reduction of the variance [20]

$$\mathrm{var}\{\overset{w}{P}(f_b)\} \approx \frac{11}{18}\frac{P^2(f_b)}{K}$$

Where K=N/L with N is length of the signal and L the window width.

Spectrogram

The respiratory sound signals were plotted in time domain or time-frequency (TF) domain to track the evolution of frequency and energy as the reflection of pathology changes and visually the physicians diagnose the respiratory sound abnormalities mostly from TF domain i.e. from spectrogram. The discrete Fourier transform used as the input signal to plots a spectrogram. We applied Kaiser Window of length NFFT with sampling frequency is 10000 Hz.

Result and Analysis

Test subjects were asked to the breath deep and recording was saved for each subjects. In this study, dataset consists of respiratory sound recorded over trachea from 100 healthy and 10 pathological subjects with different degrees of airway obstruction. The test subjects have the age range from young to elder, and were having a wide range of pulmonary dysfunction such as asthma Analysis of RS for the identification and extraction of various conditions, the variety in the associated pathology was considered.

The recorded respiratory sound signals were first filtering with band pass filter into individual inspiratory/expiratory segments, and manual classification results for these segments were obtained, respectively, from doctors in Amravati, by listening to each individual RS segment. The sound power depends on the flow of air during inspiration and expiration. The maximum flow level of a respiration cycle tends to vary due to abnormalities in breathing pattern, such as variations in respiration depth and respiration frequency.

Figure 1 show that RS signal of normal healthy patient and Figure 2 shows that RS signal of asthmatic patient. Figure 1a and 1b represent the normal respiratory sound (Bronchi Sound) in time domain and time-frequency domain in which inspiration much louder than expiration. Figure 1c shows that the frequency spectrum in which the RS signal frequencies are present below the 1500 Hz and Figure 1d shows that the power spectrum estimation having most of the peak frequencies present in normal RS sound having 600 Hz.

Figure 2 Shows that RS signal of asthmatic patient, Figure 2a and

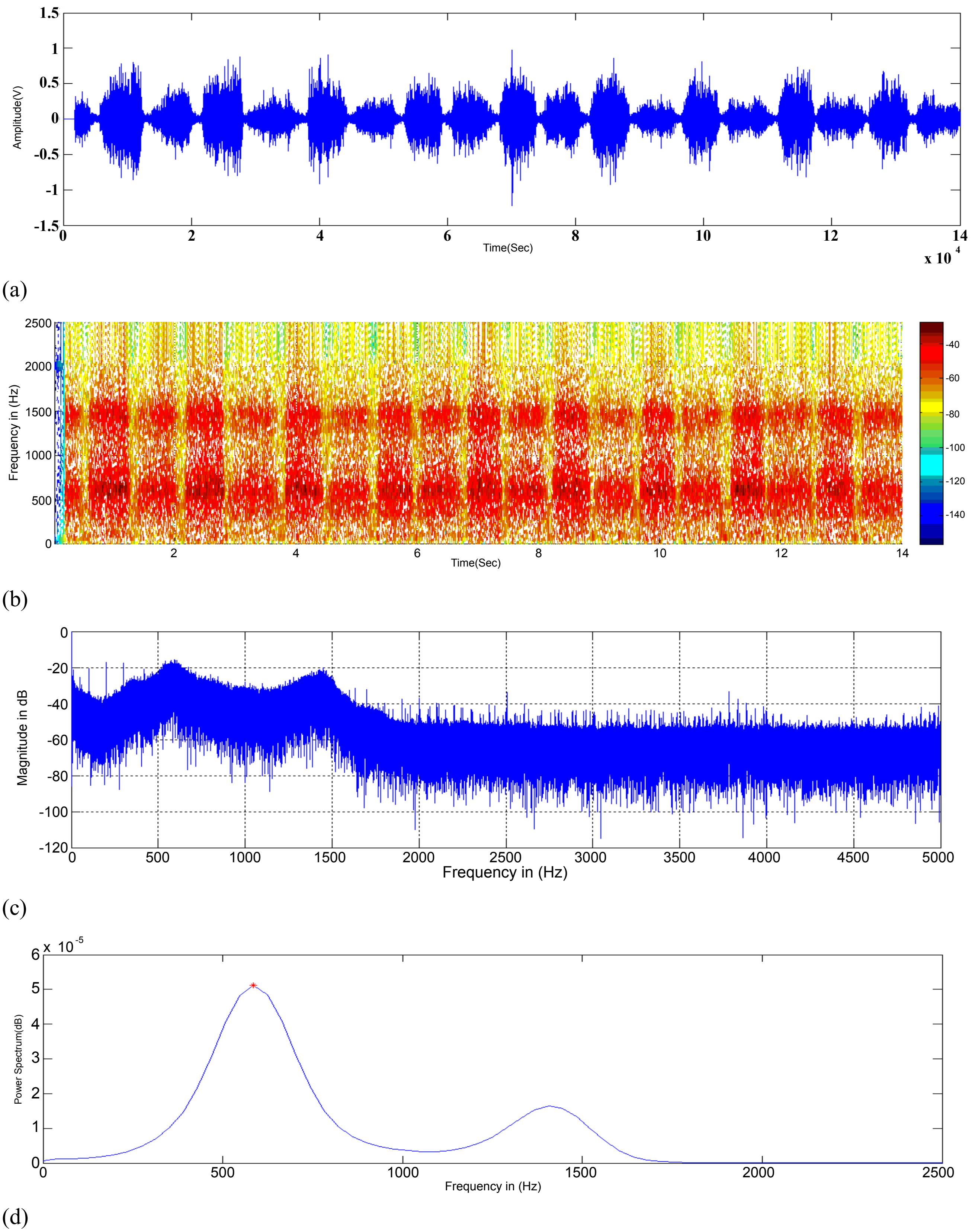

Figure 1: Respiratory sound of normal patient presented in (a) time domain, (b) Spectrogram (c) Frequency spectrum using a Kaiser window (d) Spectral Power estimation by Welch method.

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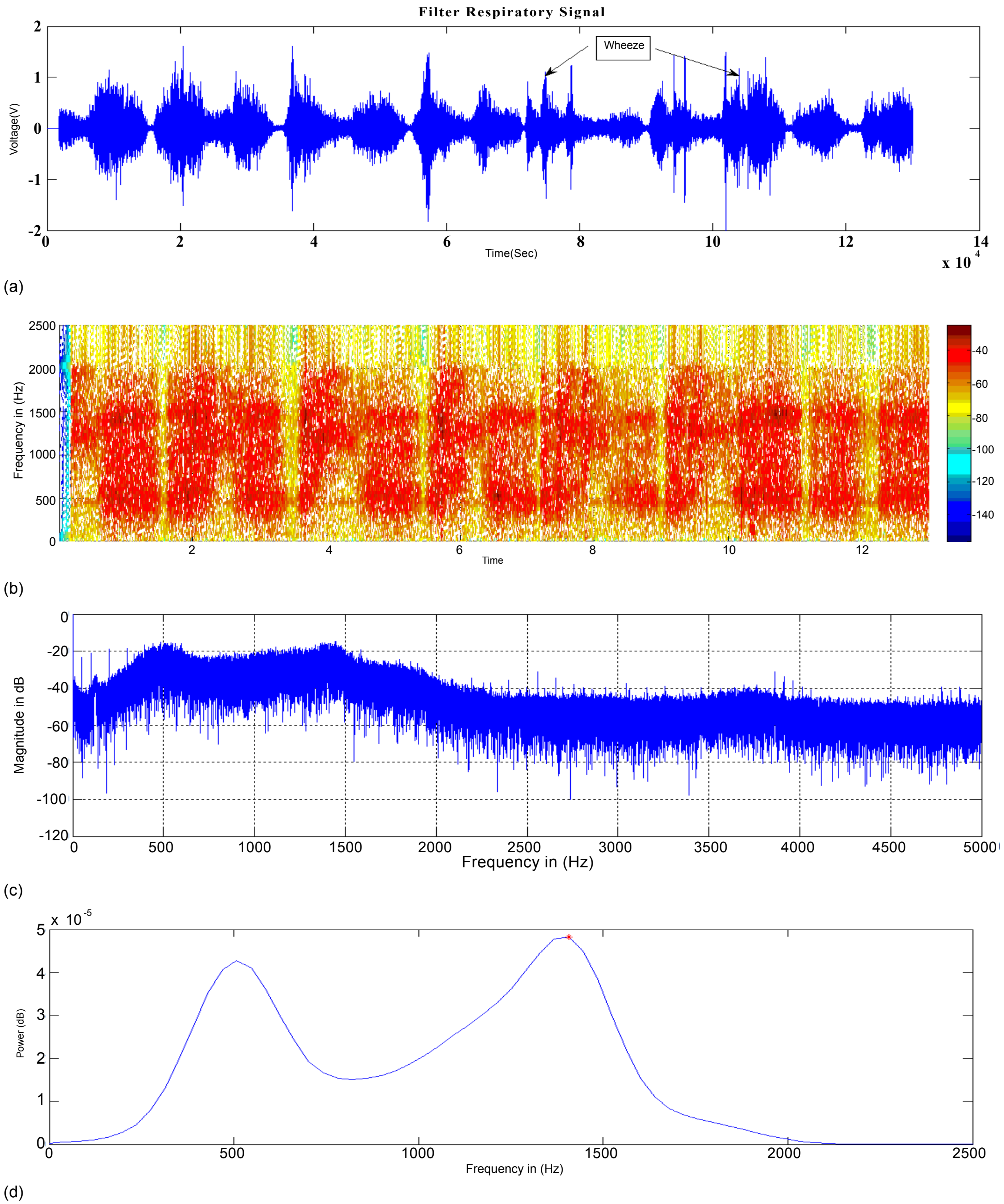

Figure 2: Respiratory sound of Asthmatic patient presented in (a) time domain, (b) Spectrogram (c) Frequency spectrum using a Kaiser window (d) Spectral Power estimation by Welch method.

b represent the wheeze present in RS signal in time domain and time-frequency domain. Figure 2c shows that the frequency spectrum in which the RS signal frequencies are present below the 2000 Hz and Figure 2d shows that the power spectrum estimation having most of the peak frequencies present in asthmatic RS signal having 1410 Hz because of change in pathological condition such as airway obstruction and flow limitation.

Table 1 show the peak frequency and average duration of wheezing, as measured from successive power spectra with time resolution 50 ms. Wheezing was detected during breathing as a result of pulmonary obstruction was declared by the physician. As Table 1 shows that the normal subject peak frequency presents in their RS signal ranging from 470 Hz to 600 Hz and in case of wheezing peak frequency components present in RS signal ranging from 1370 to 1410.

Automatic Detection System

The automatic wheeze detection system based on real-time processing and divided into hardware and software parts. The hardware part includes a microphone with amplifier, a data acquisition (DAQ) board, and a computer. Microphone and amplifier have wide frequency ranging from 50 Hz up to 10 kHz. Sixteen-bit DAQ board for converting analog signal into suitable digital signal with sampling rate of 10000 Hz are applied. The specification of the used computer as follows: Intel(R) Core(TM) i3 CPU 2.40 GHz with 3 GB RAM was employed. The algorithm for real-time detection of a wheeze sound coded with MATLAB having data acquisition and signal toolboxes.

Discussion

The works carried out in the past have concentrated more in developing wheeze sound analysis system rather than developing an automatic system that can monitor a wheeze sound in real time detect an symptom. In the above methods in current research the respiratory sound signals were plotted in time domain or time-frequency (TF) domain and visually the physicians diagnose the respiratory sound abnormalities mostly from TF domain i.e. from spectrogram. The TF waveforms of the respiratory signals are monitored to detect disorders. The disorder is identified by the frequency intensity of the signals. This type of system mainly depends on the expertise of the physicians. Visual analysis requires well trained professionals to diagnose abnormalities in the respiratory sounds. As visual analysis is purely based on the expertise of the physician, it has a high possibility of human error.

The purpose of this study is to develop an automatic system based on classical method of spectral analysis for analysing and visualizing asthmatic respiration sound can monitor a health condition in real time using acoustical information and detect an abnormal symptom.

Case	Category	Sex	Age	BMI	Peak frequency (f_p) hz
1	Healthy	M	21	19.6	600
2	Healthy	M	22	19.5	430
3	Healthy	M	40	19.7	430
4	Healthy	M	34	19.5	470
5	Healthy	F	22	19.2	470
6	Healthy	F	22	19.3	590
7	Healthy	F	21	19.2	510
8	Asthmatic	M	22	19.7	1370
9	Asthmatic	F	22	19.4	1370
10	Asthmatic	F	21	19.3	1410

Notes: Abbreviations: BMI, Body Mass Index

Table 1: Classification performance of the proposed spectral power estimation method from Datasets.

Conclusion

An algorithm to detect wheezes in the time-frequency domain and spectral power estimation, with high sensitivity, developed and validate from doctors by listening to each individual RS segment. In RS signal contain wheeze carry enough information of airways obstruction in individual asthmatic patients because of this obstruction frequencies components presents in RS signal are higher which easily identified in power spectrum. The analysis of spectral data by applying a spectral power estimation to detect the wheeze prove to be beneficial reliable, fast and accurate wheeze detection method which is essential for automatic diagnosis and treatment of respiratory disorder such as asthma.

The results indicate very significant differences between RS signal of the normal healthy person and asthmatic patient. For frequency and other analysed parameters, differences were also significant the peak frequency components present in normal subject range below 600 Hz and asthmatic patient these peak frequencies are much higher above 1300 Hz.

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Conflicts of Interest

The authors report that they have no conflicts of interest to disclose. The institute had no role in the decision to conduct this study and were not involved in the writing of the manuscript, or in the analysis or interpretation of these results.

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