Classification of Respiratory Diseases Using Respiratory Sound Analysis

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***Abstract*—Respiratory or lung sounds recorded on the chest can be used to identify different types of diseases. These sounds are attenuated by the thorax and thorax-microphone interface. In order to proper classification of respiratory diseases waveforms similar to the ones generated within the lungs must be recovered from the attenuated sounds. The equalization of crackle sounds recorded on the chest can be done for accurate classification of respiratory sounds. From an experiment an estimation of the channel attenuation was obtained according to which the equalization is applied. For that, multiple tones between 100 and 1200 Hz were applied to each subjects' mouth where they were acquired. These tones were also recorded on the chest. The power ratio between the one measured on the chest and that measured at the mouth is used to calculate the attenuation of each tone. After obtaining the average attenuation curve a discrete-time equalizer was applied to crackles acquired from patients with congestive heart failure, fibrosis, and pneumonia. The equalization is used to modify the maximum frequency and two cycle duration indices measured from these crackles. The equalizer improves the extraction of features from the crackles sounds. Equalization of crackles can be used to better classify the different diseases.**

***Index Terms*—equalization of crackles, lung diseases, respiratory disease classification, respiratory diseases, respiratory sound analysis**

I. Introduction

Respiratory sounds can be recorded with the help of devices having different technical specifications. The European Respiratory Society proposed the computerized respiratory sound analysis (CORSA) guidelines for research and clinical practice [1]. Still the characterization of respiratory sound is not accurate. The attenuation of the sounds traveling from the lungs to the thorax surface provides the crackles that can be best heard [2]. The lung sounds referred as crackles are useful for classifying cardiopulmonary diseases such as fibrosis, congestive heart failure, and pneumonia [3], [4]. The crackles are usually heard on the chest with a stethoscope during patient checkups; their identification depends on the experience and hearing perception of the physician [3]. The visual inspection of crackles recorded waveform reveals an initial fast-rising deflection followed by a short ringing duration [3], [4]. The crackle can be described as short, explosive, and transient. The quantitative characterization of crackles can be done for identification of various diseases. Various electronic systems are used to record these respiratory sounds. The diseases can be identified by the two cycle duration (2CD) index (time from the beginning of the initial deflection of a crackle to the point where the waveform of the crackle has completed two cycles) and the maximum frequency of crackles [3], [4]-[7].

The parameters that are measured from the crackles can get affected as attenuation path may modify the crackle waveform. Further information for assisting the diagnosis of the different diseases can be acquired from the crackles with characteristics closer to those generated within the lungs. The acquired sounds can be equalized by knowing the transmission channel to recover characteristics that were changed during their propagation through the path. This paper presents discrete-time equalization method for compensating sound attenuation measurements of the channel consisting of the thorax and the thorax interfaces.

The importance of listening to and understanding respiratory sounds is evident from the iconic and symbolic usage of the stethoscope in modern medicine. The stethoscope was invented in 1821 by the French Physician, Laennec, upon the discovery that respiratory sound analysis aids in the diagnosis of pulmonary infections and diseases, such as acute bronchitis and pneumonia [8], [9]. Since 1821, stethoscopes have become the most common diagnostic tool by doctors in the twenty-first century [9], [10]. Despite its widespread use, however, analysis of respiratory sounds using stethoscopes is rudimentary at best and requires a degree of subjectivity from the physician [9]-[11].

Analysis of respiratory sounds using stethoscopes depends on the variable factors of the diagnosing physician's experiences, hearing, and ability to recognize and differentiate patterns [10]. In addition, stethoscope data is not typically recordable, making long-term correlation of data difficult [9], [11]. All of these factors reduce the value stethoscopes bring to a world that increasingly demands quantitative measures of disease.

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Over the last four decades, researchers have made significant progress in fine-tuning computerized signal processing techniques, and it is now possible to perform respiratory sound analysis using many of these techniques [10]. Computerized analysis has succeeded in producing graphical representations of respiratory signals, which provide physicians additional methods of pulmonary diagnosis [10]. The use of computational power to analyze pulmonary spectral data has advanced respiratory sound analysis from a subjective skill to an objective one [9], [12]. In addition, the recent availability of cheap computer memory has enabled permanent storage of recorded respiratory sounds [12]. The next step in computerized analysis of respiratory signals is to automate the classification of respiratory sounds based on real-time data.

Respiratory signals can be classified into two major categories: normal lung sounds (NLS) and abnormal lung sounds (ALS) [13]. Most abnormal lung sounds are both adventitious and nonstationary. While many types of abnormal lung sounds exist, the two major categories of abnormal lung sounds are wheezes and crackles [13].

A wheeze is a continuous adventitious sound that is characteristically “musical” in Nature [14]. Wheezing is usually caused by airway obstruction in the lungs [13]. The presence of wheezes during breathing can indicate asthma, cyctic fibrosis, and bronchitis in a patient [13], [14]. Wheezes are high-pitched in relation to normal breath sounds, and their frequency distribution is usually between the 400 Hz to 600 Hz range [13]. They typically last for longer than 100ms [14]. Because wheezes have a defined frequency range, frequency domain analysis of a respiratory signal can reveal a wheeze.

A crackle is a discontinuous adventitious sound that is characterized by sharp bursts of energy [14]. Their duration is typically shorter than 20ms, and they are characterized by a wide distribution of frequencies [14]. Because of this wide frequency distribution, it is difficult to pinpoint crackles in the frequency domain. Crackles can be broken down into two additional categories. Fine crackles are high-pitched crackles that occur repeatedly over inspiration across multiple breathing cycles [13]. Coarse crackles are low-pitched sounds that appear early during inspiration or sometimes during expiration as a result of liquid filling small airways in the lungs [13]. The presence of crackles can indicate cardio respiratory diseases, pneumonia, and chronic bronchitis [14].

Over the last thirty years, various methods of computerized respiratory sound analysis have attempted to distinguish and classify abnormal lung sounds. These methods include the usage of Fast Fourier Transforms, Short Time Fourier Transforms, Wavelet Transforms, fuzzy logic classification, autoregressive modeling and neural network classification—among others. The study of computerized detection methods for abnormal lung sounds continues to grow as computer classification algorithms become more sophisticated over time. The detection of abnormal lung sounds in respiratory signals using the Fast Fourier Transform and the Wavelet Transform in conjunction with neural network classification.

II. RELATED WORK

In past recent different researches has been carried out to measure the thorax attenuation imposed to the lung sounds. In [2] the white noise is applied to the mouth of eight healthy individuals and measured the received power on the chest to characterize the attenuation from 100 to 600Hz. A similar procedure to measure the path attenuation from five subjects is used in [15]; instead of white noise tones from 20 to 600Hz are applied. The accelerometers are used to measure the attenuation in these studies. The acquisition of lung sounds using piezoelectric microphones or electrets microphones (capacitive microphone with a pre-charged, nonconductive membrane between the capacitor plates) housed into acoustic couplers [3] were recommended in the CORSA guidelines. The lung sounds are further modified by the thorax-microphone interface using acoustic couplers. Measurements were carried out in eight healthy subjects (seven men; age: 35 ± 5.3 years; body mass index: 26.1 ± 1.8kg/m^2*)* based on the method proposed in [15] to obtain the sound attenuation curve of the thorax and thorax-microphone interface recommended by the CORSA guidelines.

Sinusoidal tones being simultaneously recorded at the application point and on the chest as well as were applied to the subjects’ mouths. The power ratio between the one measured on the chest and that measured at the mouth was used to calculate tone attenuation. The experimental setup to measure the attenuation consists of various components. An audio editor running on a computer was use to generate twelve sinusoidal tones ranging from 100 to 1200Hz (increments of 100Hz). The amplified computer audio output was applied to a speaker with the characteristics: 4-cm diameter; maximum power of 100mW; flat frequency response from 20 to 20,000Hz; and sensitivity of −96dB (±3dB). The speaker was placed in a polystyrene box filled with attenuating material to prevent the propagation of the generated tones to the environment.

The tones were conveyed from the box to the subjects’ mouths by means of a plastic duct. An electrets microphone MD9745APA-1, Knowles Acoustics was placed at the end of this duct to record the tones applied to the mouth; their power was used as reference values for the attenuation measurements. Second electrets microphone MD9745APA-1, Knowles Acoustics housed into a nylon acoustic coupler with the dimensions recommended by the CORSA guidelines [1], was used to acquire the tones on the thorax. The acoustic coupler was fixed with double-sided tape on the left posterior chest wall close to the fifth thoracic vertebrae at a distance of 5cm from the spine. The employed microphone model has a flat frequency response from 100 to 3000Hz and sensitivity of 9mV/Pa. The sounds were recorded during the inspiratory phase when the glottis is opened. The opened glottis allows the sound propagation through the respiratory tract such that it reaches the thorax surface with higher intensity. A pneumotachograph is a device that measures flow in terms of the proportional pressure drop across a resistance. It was assembled in the duct that conveys the tones to the subjects' mouths to identify the inspiratory phase. The pressure drop across the flow resistance was measured with a differential pressure transducer DC030NDC4 - Honeywell Inc.

The subjects breathed through the microphone and the pneumotachograph during the sampling of the tones. A description of the system used to record the waveforms is given in [16]. Some of its characteristics were modified to circumvent the difficulties involved in obtaining the measurements described in [17]. The tone acquired by the microphone placed on the chest contains two circuits; having a second order Butterworth high-pass filter (HPF) and a second-order Butterworth low-pass filter (LPF). The high-frequency components of normal respiratory sounds are more attenuated by the transmission channel than the applied lower frequency tones. The wide bandwidth circuit (WBC) and the narrow bandwidth circuit (NBC) was used to measure sound from which attenuation curve was obtained.

All waveforms were simultaneously sampled at the rate of 10 kSamples/s by a quad sample-and-hold integrated circuit; next, the four sampled signals were sequentially supplied, by means of a multiplexer, to a 12-bit A/D. The digital samples were transmitted to a computer via USB interface where they were stored as individual wave files.

The sounds recorded with the WBC were used to compute the power of the tones ranging from 100 to 300Hz. The power of the other tones was measured using the recordings acquired with the NBC. The relative attenuation of the different tones by the two circuits was measured and was taken into account when calculating the attenuation curve. This approach was adopted because the gains and the HPF cutoff frequency (HPFCF). It would have been possible to use one acquisition channel to record the sounds on the chest; for that, its gain and HPFCF would have to be manually changed by means of potentiometers to avoid the excessive attenuation of a given tone. This procedure would have increased the measurement errors since it would have been difficult to reproduce the potentiometers adjustments during the measurements carried out on the different subjects. Approximately, 2s of the tones recorded at the mouth and on the chest during the opening of the glottis were used to calculate the attenuation curve. Each of these recordings was divided in segments of 512 points with each segment containing 50% of the previous segment samples to minimize the effect of spurious respiratory noises [18]. The discrete Fourier transform of each segment was calculated by applying a Hamming window to them.

III. Discrete Time Equalization for Respiratory Sound Recovery

A discrete time equalizer can be used to recover sounds with characteristics closer to the ones generated by the lungs, after measuring the attenuation of the transmission channel. It can be assumed that the thorax and the thorax-microphone interface behave as a linear and time invariant system with the respective unit impulse responses $t[n]$ and $c[n]$, the output signal $x[n]$ acquired with the microphone on the chest is given by

$$x[n] = d[n] * t[n] * c[n] = d[n] * g[n]$$

where * is the convolution operator and $d[n]$ stands for the respiratory sound.

The purpose of the equalizer is to cancel the effect of $g[n]$ in order to recover $d[n]$, such that $y[n]\approx d[n]$

$$y[n] = d[n] * g[n] * h[n]$$

where $h[n]$ is the equalizer's unit impulse response.

To cancel this effect, the convolution of the unit impulse responses of the equalizer and channel has to approximate to the unit sample

$$h[n] * g[n] \approx \delta[n]$$

which gives $y[n]\approx d[n]$. Thus, the input signal $d[n]$ can be recovered by a linear inverse filtering operation which is also referred as equalization, also called deconvolution.

To improve system performances in different fields, such as spectroscopy [19] and communications systems [20] deconvolution has been used. From above equation, $H(z)G(z) \approx 1$ and $H(z) \approx 1/G(z)$, where $H(z)$ and $G(z)$ are the z transforms of $h[n]$ and $g[n]$, respectively. In order to obtain a $H(z)$ that corresponds to a causal and stable system, it is necessary that $G(z)$ has minimum phase [14].

MATLAB is used to sample the magnitude response of a FIR filter that imitates the measured attenuation curve ($G(e^{j\omega})$). The inverse of FIR filter ($H(e^{j\omega})$) contained poles outside the circle of unit radius. FIR least square inverse (FLSI) method [21] is an alternative approach to obtain a stable $h[n]$. The performance of the $h[n]$ obtained with the FLSI method depends on the choice of its number of coefficients (N). The effect of N on the equalizer response can be assessed by applying $h[n]$ of different sizes to crackles acquired from stable ambulatory fibrosis patients. In order to compare the equalized crackle to the applied one, they were normalized by their respective maximum amplitudes.

A. Maximum Frequency and Two Cycle Duration Measurement

The discrete time equalizer is used to assess the impact of the equalization on indices measured from the crackles acquired from patients with pulmonary fibrosis, heart failure, and pneumonia. Crackles were recorded from patients for each disease. These crackles were obtained from [22] in which their maximum frequencies and 2CD indices without equalization were measured [22]. The patients, guided by a metronome, were instructed to breathe through a duct containing a pneumotachograph at the rate of 12 breaths/min. The crackles and the flow waveform were acquired at the rate of 10 kSamples/s with the system developed in [18].

The flow waveform allows for the localization of the crackles within the inspiratory cycle. The crackles were identified by visual inspection of the recordings using the audio editor. Each crackle was stored into a separated file and all them were organized according to the disease and patient identification number. Four crackles within the same inspiratory phase were selected for each patient. This was repeated for different patients diagnosed with the same disease. The average maximum frequency and

2CD index per patient were calculated. Their maximum frequency was estimated by applying Wigner–Ville distribution and the modified geometric method [22], [23]. From those measurements, the mean and the standard deviation of the maximum frequency and 2CD were calculated for each disease. In this paper, the same sets of crackles were equalized and then, the measurements of their maximum frequencies and 2CD indices were taken with the same techniques used in [22].

The parameters measured for each illness have normal distribution according to the Shapiro–Wilk test [24]. The indices obtained before equalization (BE) and after equalization (AE) were analyzed with the one-way ANOVA followed by Newman–Keuls test for comparison among groups [25]. *P* values <0.05 were considered statistically significant.

IV. Experimental Results and Discussion

The average attenuation values and its standard deviation are obtained from various respiratory sounds. The transmission channel acts as a LPF. Therefore, data on lung sounds with broad spectra will be distorted by the channel. Thus, the application of an equalizer may recover data from the lung sounds allowing the extraction of better quantitative diagnosis indices. In order to find out the most suitable filter length (*N*) for the proposed equalizer, crackle waveforms applied to the subjects' mouths were normalized by their respective maximum amplitudes; the same crackles were recorded on the chest, equalized by filters designed with different number of coefficients, and normalized by their respective maximum amplitudes. Afterward, the mean absolute error (MAE) and the root mean square error (RMSE) of the different equalized crackles were calculated

$$\text{MAE} = \frac{1}{M}\sum_{i=i}^{M} \; |y_i - \overline{y_i}|$$

$$\text{RMSE} = \sqrt{\frac{1}{M}\sum_{i=i}^{M} \; (y_i - \overline{y_i})^{\,2}}$$

where y_i is the normalized sample value of the crackle recorded at the mouth; $\overline{y_i}$is the normalized sample value of the equalized crackle recorded on the chest, and M is the number of crackle samples. The crackles recorded on the chest of each individual were reasonably corrupted by the channel.

TABLE I. MAE and RMSE Values Calculated between the Crackle Waveforms Sampled in the Mouth and on the Chest

Classification	MAE_{BE}	$RMSE_{BE}$	MAE_{AE}	$RMSE_{AE}$
Corrupted Crackle	0.18	0.23	0.12	0.15
Very Corrupted Crackle	0.28	0.34	0.13	0.18

Table I shows the MAE and RMSE calculated before and after the equalization by filters of different sizes applied to the crackles acquired on the chest.

Table II shows the mean values and standard deviation of the maximum frequency and 2CD index measurements for crackles before and after the equalization. This table also reproduces the measurements of [22] in which the crackles were not equalized. Equalization was carried out with the designed 26-coefficient FIR filter. Between-groups comparisons were carried out for the 2CD and maximum frequency indices. A summary of significance values for all comparisons is given in Table III. For both sets of crackles, BE and AE, the statistical analysis showed that the mean value of the maximum frequency and 2CD are significantly different between fibrosis and pneumonia crackles, and also between fibrosis and heart failure crackles (p<0.05). Similarly, for both sets, there are no statistical significant differences between the mean values of the parameters for pneumonia and heart failure.

TABLE II. The Maximum Frequency and Two Cycle Duration Indices Obtained before Equalization and after Equalization.

Disease Classification	Non-Equalized Crackles		Equalized Crackles	
	Maximum Frequency (Hz)	2CD (ms)	Maximum Frequency (Hz)	2CD (ms)
Fibrosis	904.32±157.32	4.87±0.66	1098.47±145.54	3.38±0.99
Pneumonia	704.29±265.35	6.58±2.04	955.89±260.43	4.17±1.13
Heart Failure	640.22±153.23	7.05±1.52	898.71±171.82	4.04±0.87

TABLE III. Comparison of Maximum Frequency and Two Cycle Durations on the Basis of Significance Values (p).

Diseases	Maximum Frequencies		Two Cycle Duration	
	BE	AE	BE	AE
Fibrosis/ Heart Failure	0.006	0.007	0.027	0.018
Fibrosis/ Pneumonia	0.018	0.011	0.011	0.003
Pneumonia/ Heart Failure	0.352	0.892	0.778	0.069

V. Conclusion and Future Scope

This paper presented the discrete equalization method to compensate sound attenuation enforced by the thorax and thorax-microphone interface with a system based on the CORSA guidelines. The diseases are classified after analyzing the recorded respiratory sounds. From the sound attenuation measurements obtained from healthy individuals, it is possible to observe that lung sounds with broad spectra, such as crackles, have their frequency components unevenly attenuated. Therefore, the sounds recorded on the chest are modified during their propagation through the thorax, preventing the extraction of consistent diagnosis indices from the crackles. The disease classification is improved after respiratory sound equalization.

In order to circumvent the inter individual variability of the thorax sound impedance that may also change for different diseases and diseases stages, more elaborate equalization techniques need to be investigated in near future. The disease classification results can be further improved by using neuro fuzzy systems.

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