40th Annual Conference of the International Lung Sounds Association

Conference Proceedings

CD

St. Petersburg, Russia, 24th – 25th September 2015

PROCEEDINGS

of the 40th International Lung Sounds Association Conference

St. Petersburg Electrotechnical University “LETI”

St. Petersburg, Russia

24th – 25th September 2015

OPENING REMARKS AND WELCOME MESSAGES TO THE 40th ANNUAL ILSA CONFERENCE

Dear Friends and Colleagues,

It is a great honor to address to you from across the world, I am an eccentric, Japanese-American, named Sadamu Ishikawa, M.D., FCCP, from Boston, MA, as I'm current President of ILSA.

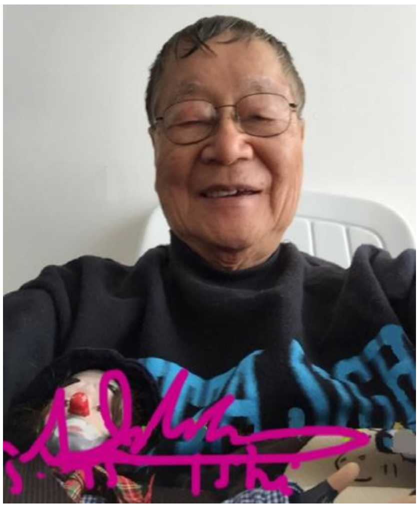
Drs. Robert Loudon and Raymond L. H. Murphy, Jr conceived idea of having a scientific meeting of 'Auscultation'. In October 1976, International Lung Sounds Association (ILSA, http://www.ilsaus.com) was founded and 1st ILSA meeting was held in Boston, MA. Following year after year, this annual meeting of ILSA continued, without interruption till last year, October, 10th&11th, 2014, in Boston, MA, USA. Achievement of these 40 years of annual meetings of ILSA, without any interruption has been accomplished by my great teacher Raymond, L.H. Murphy, Jr. Following this year's September 24th & 25th, 2015 - 40th ILSA at Saint Petersburg, Russia, we'll have 41st ILSA on October 7th & 8th, 2016, in Tokyo, Japan, then 42nd, ILSA in the Fall, 2017 in Istanbul,Turkey,then in 2018, 43rd ILSA to be held in Los Angels, California, U.S.A.

Auscultation and respiratory acoustics have many features. Dr. Raymond Murphy, repeated Laennec's original experiment, namely Producing Crackles on Popping Salts in Hot Frying Pan and Auscultated the Crackles.

I believe that auscultation should remain as a main part of Physical Examination for years to come, so as Annual Meetings of ILSA.

My best wishes to this annual conference of International Lung Sounds Association.

President of ILSA
Sadamu Ishikawa,
M.D., FCCP,
Boston, MA, USA



OPENING REMARKS AND WELCOME MESSAGES TO THE 40th ANNUAL ILSA CONFERENCE

We are very happy to finally welcome this 40-th Annual ILSA meeting in St. Petersburg, Russia. This meeting is the 1-st ILSA Conference in Russia. For this reason we’d like to introduce to the Respiratory Acoustic Community from the entire world some achievements made in Russia and former USSR and mostly published in Russian.

The first studies on Respiratory Acoustics in Russia (former USSR) were performed in the late 1970-s by Leonid Nemerovsky and coauthors. In 1980-th G. Lyubimov and A. Dyachenko investigated biomechanical and acoustical properties of the lungs. In the late 1980-s and early 1990-s the conversion of the defense studies into civilian areas resulted in transition of many new experienced researchers from general acoustics into respiratory acoustics. Many important theoretical and experimental results were achieved since that time in Kiev in the Institute of Hydromechanics of National Academy of Sciences of Ukraine (www.hydromech.com.ua) by V. Grinchenko, I.Vovk et al. In the late 90-s another group emerged in Vladivostok (V. Korenbaum, A. Tagiltsev, Yu. Kulakov et al.).

In the 21-st century Russian respiratory acoustics researchers participate more in the national and international periodic and established more contacts. The periodic includes Russian Journals, translated and published in English as Human Physiology, Acoustical Physics, Fluid Dynamics and high-impact international journals like J. Biomechanics, Respirology. They presented reports at the meetings of Acoustical Society of America, European Respiratory Society, American Thoracic Society, Congresses on Biomechanics and Biomedical Engineering. Since 2007 year the Russians participate in Annual Meetings of ILSA. And now we meet 40th ILSA Conference in Russia.

Research activity in respiratory acoustics changed with time, increasing and decreasing. The last decrease in the first decade of 21-th century was induced by hopes that CT, MRI, PET will visualize the lungs and physicians will get the necessary information. But now they understand more and more that the techniques provide structural but not functional information. For this reason interest to the acoustical studies of respiratory system once more increases. We hope that 40-th jubilee ILSA meeting will be an important milestone on this ascent of respiratory acoustics.

Acknowledgments

At first we’d like to thank the secretary of the Organizing Committee of ILSA-40 miss Anna Glazova. From the very beginning she managed the complete business of the preparation for the Conference. We got a worm support in LETI from Prof. Michael Shestapalov and others. Students of the Bauman Moscow State Technical University Ekaterina Fomina and Maria Veremyeva were very helpful in preparing this book of abstracts.

For the local organizers

Alexander Dyachenko

Vladimir Korenbaum

LIST OF ILSA CONFERENCES

No.	Date	Place	Local Organizer(s)
1.	October 1976	Boston, MA	Raymond L.H.Murphy,Jr.
2.	September 1977	Cincinnati, OH	Robert Loudon
3.	September 1978	New Orleans, LA	William Waring
4.	September 1979	Chicago, IL	David Cugell
5.	September 1980	London, England	Leslie Capel & Paul Forgacs
6.	October 1981	Boston, MA	Raymond L.H.Murphy,Jr.
7.	October 1982	Martinez, CA	Peter Krumpe
8.	September 1983	Baltimore, MD	Wilmot Ball
9.	September 1984	Cincinnati, OH	Robert Loudon
10.	September 1985	Tokyo, Japan	Riichiro Mikami
11.	September 1986	Lexington, KY	Steve S. Kraman
12.	September 1987	Paris, France	Gerard Charbonneau
13.	September 1988	Chicago, IL	David Cugell
14.	September 1989	Winnipeg, Canada	Hans Pasterkamp
15.	October 1990	New Orleans, LA	David Rice
16.	September 1991	Veruno, Italy	Filiberto Dalmasso
17.	August 1992	Helsinki, Finland	Anssi Sovijärvi
18.	August 1993	Alberta, Canada	Raphael Beck
19.	September 1994	Haifa, Israel	Noam Gavriely
20.	October 1995	Long Beach, CA	Christopher Druzgalski
21.	September 1996	Chester, England	John Earis
22.	October 1997	Tokyo, Japan	Masahi Mori
23.	October 1998	Boston, MA	Sadamu Ishikawa
24.	October 1999	Marburg,Germany	Peter von Wichert
25.	September 2000	Chicago, IL	David Cugell
26.	September 2001	Berlin, Germany	Hans Pasterkamp
27.	September 2002	Helsinki, Stockholm	Anssi Sovijärvi
28.	September2003	Cancun, Mexico	Sonia Charleston, Ramyn Gonzales Camarena & Tomбs Aljama Corrales
29.	September 2004	Glasgow, Scotland	Ken Anderson & John Earis
30.	September 2005	Boston/Cambridge, MA	Raymond L.H.Murphy,Jr.
31.	September 2006	Halkidiki, Greece	Leontios Hadjileontiadis
32.	November 2007	Tokyo, Japan	Shoji Kudoh
33.	October 2008	Boston, MA	Sadamu Ishikawa & Raymond L.R. Murphy,Jr.
34.	September 2009	Haifa, Israel	Noam Gavriely
35.	October 2010	Toledo,OH	Dan E. Olson
36.	September 2011	Manchester, UK	Ashley Woodcock
37.	October 2012	Rochester, Minnesota	Michael E. Nemergut
38.	November 2013	Kyoto, Japan	Yukio Nagasaka
39.	October 2014	Boston, MA	Sadamu Ishikawa
40.	September 2015	St. Petersburg, Russia	Alexander Dyachenko, Vladimir Korenbaum & Zafar Yuldashev

GENERAL INFORMATION

Conference Venue/Accommodation

St. Petersburg Electrotechnical University “LETI” (http://eltech.ru/en/university), ul. Professora Popova 5, 197376 St. Petersburg, Russian Federation

Official language: English

Supporting organizations:

Saint-Petersburg Electrotechnical University “LETI”
State Scientific Center of the Russian Federation – Institute of Biomedical Problems of the Russian Academy of Sciences
V.I. Il'ichev Pacific Oceanological Institute of the Russian Academy of Sciences

Organizing Committee

Vladimir Kutuzov – Co-Chair, , Prof., Dr.Sc., Rector of St. Petersburg Electrotechnical University “LETI”, St. Petersburg, Russia
Igor Ushakov – Co-Chair, Prof., Academician, Director of the Institute of Biomedical Problems of RAS, Moscow, Russia
Alexander Dyachenko – Prof., Dr.Sc., General Physics Institute & Institute of Biomedical Problems of RAS, Moscow, Russia
Vladimir Korenbaum – Prof., Dr.Sc., V.I. Il'ichev Pacific Oceanological Institute, Vladivostok, Russia
Zafar Yuldashev – Prof., Dr.Sc., St. Petersburg Electrotechnical University “LETI”, St. Petersburg, Russia
Sadamu Ishikawa – MD, President of ILSA, Boston, USA
Noam Gavriely – MD, Prof., Technion -Israel Institute of Technology, Haifa, Israel
Anna Glazova – Secretary of the Committee, St. Petersburg Electrotechnical University “LETI”, St. Petersburg, Russia

Program Committee

Sadamu Ishikawa – Chair, President of ILSA, Boston, USA
Vladimir Korenbaum –Vladivostok, Russia
Alexander Dyachenko – Moscow, Russia
Irina Pochekutova – Vladivostok, Russia
Zafar Yuldashev – St. Petersburg, Russia
Olga Kuznetsova – St. Petersburg, Russia
Ray Murphy – Boston, USA
Shoji Kudoh – Tokyo, Japan
Yukio Nagasaka – Kyoto, Japan

Certificate of attendance: participants, duly registered, will receive certificates of attendance upon requests.

Lunch will be hold in the University dining room (5th building)

Get-together party will be hold on 24th of September at 18.00 at the University restaurant (5th building)

Sightseeing tour will be hold on 25th September, 14.40 – 19.40, meeting place **-** 5th building yard, near the monument to Alexander Popov.

The Conference is supported by the
Russian Foundation for Basic Research (RFBR),
Grant № 15-01-20696

CONFERENCE PROGRAM

24th September

<u>9:00 – 10:00 – registration</u> (5th building, conference hall).

<u>10:00 – 10:30 – opening ceremony:</u>

1. Welcome address of LETI Rector Prof. Viktor Kutuzov.
2. Welcome address of ILSA President Sadami Ishikawa.
3. Welcome address of IBMP RAS Director Academician Igor Ushakov.
4. Welcome address of ILSA-40 organizers – Alexander Dyachenko.

<u>10:40 – 13:00 – session 1 (6 reports). Chairmen: Shoji Kudo, Alexander Dyachenko</u>

10:40 – 11:00 – *Noam Gavriely, Yulia Goryachev, Yael Buchnik and Amir Ohad.*
«Measuring CABS in obstructive airways diseases».

11:00 – 11:20 - coffee break

11:20 – 11:40 – *Noam Gavriely, Yael Buchnik, Yulia Goryachev and Amir Ohad.*
«Quantitative CABS monitoring in asthma management».

11:40 –12:00 – *Irina Pochekutova, Vladimir Korenbaum, Veronika Malaeva, Anatoly Kostiv, Valentina Kudryavtseva.*
«Bronchial obstruction patients clusterization by means of two-dimensional analysis of forced expiratory tracheal noise time and band pass energy».

12:00 – 12:20 – *Anna Glazova, Anastasiya Makarenkova, Ekaterina Tur, Marina Pohaznikova.*
«Comparison of chest lung sounds in patients with asthma, COPD and healthy lungs».

12:20 – 12:40 – *Juan Carlos Aviles-Solis, Peder Halvorsen, and Hasse Melbye.*
«Inter-observer variation in categorizing lung sounds».

12:40 – 13:00 – *Yukio Nagasaka, Michiko Tsuchiya, Chizu Habukawa, Chikara Sakaguchi, and others.*
«Breath sounds as biomarker in the management of bronchial asthma».

<u>13:00 – 14:00 – lunch</u> (5th building, dining hall)

<u>14:00 – 16:20 – session 2 (8 reports). Chairmen: Noam Gavriely, Yukka Rasanen</u>

14:00 – 14:20 – *Michiko Tsuchiya, Yukio Nagasaka, Chikara Sakaguchi, and others.*
«Fine crackles as biomarker of interstitial pneumonia».

14:20 – 14:40 – *Vladimir Korenbaum, Irina Pochekutova, Veronika Malaeva and Anatoly Kostiv.*
«Acoustic biomechanical relationships of forced exhalation revealed by analysis of variance among groups with various degree of bronchial obstruction».

14:40 – 15:00 – *Veronika Malaeva, Irina Pochekutova, Anatoly Kostiv, Svetlana Shin and Vladimir Korenbaum.*
«The correlation between forced expiratory tracheal noise parameters and lung function characteristics in healthy, asthma and COPD patients».

15:00 – 15:20 – *Irina Pochekutova, Vladimir Korenbaum, Veronika Malaeva.*
«An approximate estimation of forced expiratory bronchial resistance in asthma and COPD patients by means of biomechanical and acoustical surrogate measures».

15:20 – 15:40 – *Alexander Dyachenko, Anna Mikhaylovskaya, Victor Vasiliev.*
«On the nature of elastic waves generated by chest wall percussion».

15:40 – 16:00 – *Anton Shiryaev, Vladimir Korenbaum.*
«The features of sound propagation through healthy human lungs, revealed by transmission sounding with complex acoustic signals in 80-1000 Hz frequency band».

16:00 – 16:20 – *Maria Safronova, Irina Pochekutova, Vladimir Korenbaum, Veronika Malaeva.*
«Bronchodilator response of peak frequency of forced expiratory wheezes in healthy and patients with bronchial obstruction».

16:20 – 16:40 - coffee break

16:40 – 17:40 – session 3 (Skype presentations). Chairmen: Sadamu Ishikawa, Vladimir Korenbaum

16:40 – 17:00 – *Alex Rudnitskii.*
«Heart sound cancellation from lung sound recordings using empirical mode decomposition technique».

17:00 – 17:20 – *Thomas Talbot.*
«Assessment metrics & performance specifications for a virtual standardized patient comprehensive pulmonary auscultation simulator».

17:20 – 17:40 – *Raymond Murphy, Founder of ILSA.*
«History of ILSA».

18:00 – 20:00 – get-together party (5th building, University restaurant).

25th September

9:00 – 11:40 – session 4 (8 reports). Chairmen: Zafar Yuldashev, Noam Gavriely

9:00 – 9:20 – *Masato Takase.*
«Determination of lung sound spectral parameters with and without background noise subtraction».

9:20 – 9:40 – *Vladimir Korenbaum, Alexandr Tagiltcev, Anatoly Kostiv, Sergei Gorovoy and Anton Shiryaev.*
«A comparative analysis of acoustic sensors for recording respiratory sounds at the chest surface».

9:40 – 10:00 - *Hiroshi Nakano and Akemi Nakano.*
«Development of a digital filter to convert mic-sounds to stethoscope-sounds».

10:00 – 10:20 – *Sadamu Ishikawa, and Peter La Camera.*
«Small pneumothorax managed without chest tube insertion with guide of lung sound mapping in 2 young patients with polycystic lung disease».

10:20 – 10:40 – *Sadamu Ishikawa, Peter La Camera.*
«Cardiac response to respiration in deep breath-in vs. 2 breath-in and 2 breath-out».

10:40 – 11:00 – *Alexander Kalinkin, A.N. Varaksin, S.B. Gatilov, N.I. Kurenkov.*
«New approach to evaluation of sleep breathing disorders based on the analysis of the sounds of snoring and pauses of breathing cycle».

11:00 – 11:20 – *Victor Kopiev, Michail Mironov and Michail Yakovets.*
«On a possible mechanism of sound generation and amplification in a corrugated tube».

11:20 – 11:40 – *Anatoly Kostiv, Vladimir Korenbaum.*
«Monitoring physiologic status of a diver by means of respiratory sounds recorded under diving suite in situ».

11:40 – 12:00 – coffee break

12:00 – 13:10 - session 6 (posters). Chairperson: Hiroshi Nakano, Olga Kuznetsova

1. *Anna Mikhaylovskaya, Alexander Dyachenko, Antonina Osipova.*
«Tracheal sounds of forced expiration during short- and long-term immersion».

2. *Yuri Gorshkov.*
«Computerized respiratory sounds analysis on the basis of multilevel wavelet transform».

3. *Anna Poreva, Yevgeniy Karplyuk, Anastasiia Makarenkova and Anatoliy Makarenkov.*
«Application of polyspectrum analysis to diagnostic signs' detection of lung Sounds in patients with the chronic obstructive pulmonary disease».

4. *Natalya Geppe, Elena Pavlinova, T.I. Safonova, Irina Kirshina.*
«Possibilities of a computer bronchophonography in diagnostics malfunction of external respiration at children with cystic fibrosis».

5. *Natalya Geppe, Svetlana Shatalina, Vladimir Malishev, L.S. Starostina, N.G. Kolosova, A.M. Borovkova, Buharov D.G.*
«Computer bronchophonography as a new method for lung function assessment in children with bronchopulmonary diseases».

6. *Vladimir Korenbaum, Alexandr Tagiltcev, Sergei Gorovoy, Anton Shiryaev and Anatoly Kostiv.*
«On estimating a wheezing source distance by means of intensimetry processing of sound responses recorded above the chest».

7. *Dmitry Balakin, Vitaly Shtykov.*
«Processing breath sounds with the Gauss-Hermite functions».

8. *Alexander Dyachenko, Gregory Lyubimov, Inna M. Skobeleva, Mark M. Strongin.*
«Study of sound generation in the area of dynamic trachea collapse as a source of forced expiratory tracheal noise».

13:10 – 13:40 - closing ceremony

1. From ILSA-40 organizers – Vladimir Korenbaum,
2. From the host-organizers of ILSA-40 – Zafar Yuldashev,
3. ILSA President Sadamu Ishikawa,
4. From ILSA-41 organizers – Masato Takase.
5. Wishing to speak.

13:40 – 14:40 - lunch

14:40 – 19:40 - city tour, meeting place: 5th building yard, near the monument to Alexander Popov.

Inter-observer Variation in Categorizing Lung Sounds

J.C. Aviles-Solis, P. Halvorsen and H. Melbye

General Practice Research Unit, Institute of Community Medicine, UiT, The Artic University of Norway, Tromsø, Norway.

Abstract—**The aim of this study was to measure the variation in categorizing lung sounds. We took recordings from a sample of patients and sent them to medical professionals for evaluation. We calculated Intra-class Correlation Coefficient (ICC) and kappa statistics and obtained the following: ICC for any abnormal sound: .429, Wheezes: .422 and Crackles: .419. The mean kappa of the participants was: For any abnormal sound: .631, Wheezes: .684 and Crackles: .594. We conclude that the variation for categorizing lung sounds lies within moderate to good levels in most of our analysis.**

Keywords— **Inter-observer variation, crackles, wheezes, lung auscultation.**

I. Introduction

Lung auscultation is an ancient technique and is used in the everyday clinical practice. The stethoscope itself is a symbol of medicine, but in the later years the utility of this technique has been challenged and other modern techniques preferred over it [1]. Variation in clinical examination is one of the reasons that newer methods are preferred.

The objective of this study was to measure variability in the classification of lung sounds between professionals in different circumstances and from different countries.

II. Materials and methods

We collected recordings of lung sounds in a sample of healthy subjects and patients who attended a four weeks program for lung rehabilitation in Northern Norway. The total number of subjects was 20. A wireless microphone (Sennheiser MKE 2-EW with Sennheiser wireless system EW 112-P G3-G) placed in the tube of a stethoscope (Littmann Classic) was used to make the recordings. We connected the microphone to a digital recorder (Zoom Handy recorder H4n, Zoom Corporation Japan) and obtained files in .wav format. We recorded sounds at six different locations, 3 locations on each side of the thorax (Anterior thorax: Mid-clavicular line and second rib. Posterior thorax: Between the spine and the medial border of the scapula 8 cm above the inferior angle of the scapula; At the middle of the space between the spine and the mid axillary line 8 cm above the 10^{th} rib). We asked the subjects to breathe deeply with an open mouth for 2-3 respiratory cycles.

We created spectrograms of the recordings with the help of computer software (Adobe Audition, v 5.0). We made videos of each recording showing the spectrograms and the sound simultaneously and inserted them into a power point presentation with a total of 120 cases.

This presentation was sent to general practitioners in Netherlands (n=4), England (n=4), Russia (n=4) Norway (n=4). The responded also included a group of researchers in the field of lung sounds from Portugal (n=1), Canada (n=1) and Finland (n=2), Pulmonologists at the University Hospital of Nothern Norway (n=4) and a group of sixth year students from the medical program at the Faculty of medicine, UiT the Artic University of Norway (n=4). We offered the respondents two options to send an answer: either on a written format or through a Microsoft Access based electronic format.

We calculated the variation for the group using Intra-class correlation coefficient. Then, we created a reference standard with the answers of the four lung sound researchers. We considered a positive finding when three experts agreed on it. We calculated the kappa values of every participant against this standard.

III. Results

We received responses from the 28 individuals invited to participate. Among 28 raters we obtained the following ICC coefficients (see table 1): Any abnormal sound: .429 (.366-.502 95% CI), Wheezes: .422 (.359-.495 95% CI), Crackles: .419 (.358-.491 95% CI). The mean kappa of the participants compared to the reference standard was for any abnormal sound (See figure 1): .631 (.257-.928), Wheezes: .684 (.143-1,000), Crackles: .594 (.212-.873). There was variation between groups.

Table 1: ICC values of the 28 observers on the different variables

Variable	ICC	CI 95%	P-Value
Abnormal Sounds	0,429	0,366 - 0,502	<0,001
Inspiratory Crackles	0,405	0,344 - 0,476	<0,001
Expiratory Crackles	0,302	0,249 - 0,367	<0,001
Crackles NS*	0,026	0,013 - 0,044	<0,001
Crackles Total	0,419	0,358 - 0,491	<0,001
Inspiratory Wheezes	0,132	0,100 - 0,175	<0,001
Expiratory Wheezes	0,418	0,357 - 0,489	<0,001
Wheezes NS*	0,035	0,020 - 0,055	<0,001
Wheezes Total	0,422	0,359 - 0,495	<0,001
Noise	0,050	0,033 - 0,074	<0,001
Other	0,032	0,018 - 0,051	<0,001

* Not sure of respiratory phase

IV. Discussion

We presented the results of an attempt to quantify variation on the categorization of lung sounds. We were able to put together a relatively big and variate sample for an inter-observer study. According to the categorization of kappa values [2] we found that the variation on categorizing lung sounds lies within moderate to good levels. This makes us think that the findings observed during lung auscultation are reliable enough.

We do have to say that these ratings are not directly equivalent to the evaluation of lung sounds in the everyday clinical practice. The use of spectrograms in the survey could have had an influence in the results we achieved.

V. Conclusions

We conclude that the variability of lung sound classification, even though present, lies mostly within moderate to good levels of agreement.

Acknowledgement

We would like to thank Alda Marques and Hans Pasterkamp for their inspiration and inputs.

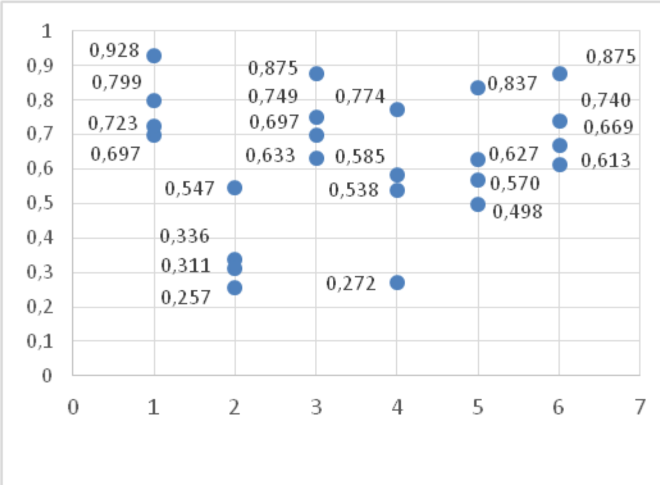


Fig. 1: Kappa values for 24 observers divided in the following groups: 1.- GP's Norway, 2.- GP's Russia. 3.- GP's UK, 4.- Pulmonologists, 5.- GP's Netherlands, 6.- Medical Students

References

[1] Howard Markel (2006) The stethoscope and the art of listening, N Engl J Med 965:325–329 DOI 10.1056/NEJMp048251

[2] Viera A, Garrett J, (2005) Understanding interobserver agreement: The kappa Statistic, Fam Med 37(5):360–3.

Processing Breath Sounds with the Gauss-Hermite Functions

D.A. Balakin, V.V. Shtykov

Moscow Power Engineering Institute Department of Fundamental Radio engineering, Moscow, Russia

Abstract – **The main principle of the acoustic signals (breath sounds, vibroacoustic noise of machine and engine) processing with the Gauss-Hermite functions is described. The processing algorithm is a modification of a wavelet transform. The parent function is formed as series of the Gauss-Hermite functions associated with an orthogonal filter bank. The signal detection is realized by means of cross-correlation function processing. As a result, the developed algorithm allows us to detect features of the signal, based on which the matched filter is constructed. The example of processing of the quiet breathing noise is described.**

Keywords – **Gauss-Hermite functions, wavelet transform, parent function, matched filter, cross-correlation function.**

I. Introduction

The analysis of the shape and parameters of acoustic signals is an integral part of the diagnosis of the condition of a wide class of objects. As a rule there is no in need simple analysis of such signals; the main purpose is identification of some of the signal's local features and period. Thus it is possible to build rythmogram of any features of the signal.

As respiratory sounds are localized in time it is reasonable to focus on the process's local features, which can not be detected by the traditional Fourier transform. A mathematical method, which known as a wavelet transform, is widely used to solve this problem [1]. The application of this mathematical apparatus allows us to identify differences in the process's characteristics at various points of time on the entire interval. Specially selected mathematical functions - parent functions or model signals are used. In addition wavelet analysis allows signals detection and selection based on a complex feature set of their distinctive shape.

The paper discusses the main principles of acoustic recordings processing on the basis of a modified wavelet transform algorithm. In contrast to the conventional approaches, we propose to use Gauss-Hermite functions (GHF) as a parent function [2].

II. Results

It is known that the Gauss-Hermite functions form a complete orthogonal system of functions in the space $L_n(-\infty,+\infty)$. From a computational point of view, one advantage of the GHF is their localization both in time and spaces of frequency [3]. Additionally, as the GHF system is mathematically full, it can be used to build a parent function of practically any shape. Therefore it will help to detect and localize a wide class of a research process's characteristics [4]. In medical practice this ability can increase the accuracy of diagnosis.

As an example considers the sample of a quiet breathing acoustic record obtained at some distance from the source (fig. 1.).

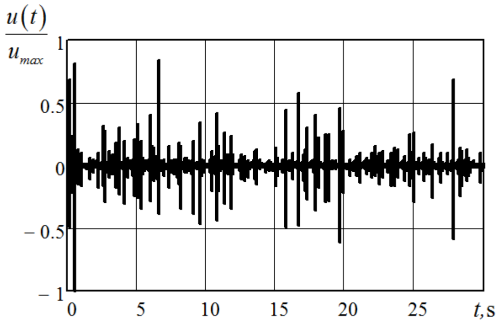


Fig. 1: Record of quiet breathing noise

After analyzing the record duplicate fragments in the form of a pair of pulses with opposite phase high frequency filling was found (Fig.2). One of these fragments was selected as the pattern for the formation of parent function.

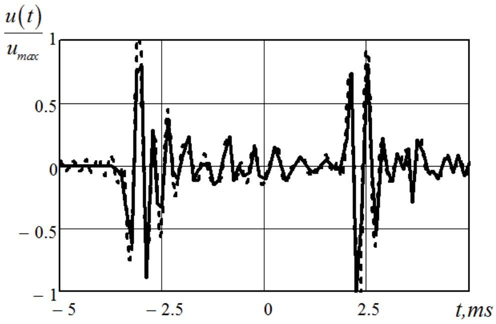


Fig. 2: Pattern (dash) and parent function (solid)

The pattern decomposition in a series of the Gauss-Hermite functions was performed. The spectrum in Hermite space was limited by value of the order of 140. Thus the following parent function form was obtained:

$$W(t,a) = \frac{\sum_{n=0}^{140} A_n(a)\varphi_n(t,a)}{\sum_{n=0}^{140} |A_n(a)|^2},$$

where a – scale factor, $A_n(a)$ – coefficients of expansion in the GHF, $\varphi_n(t,a)$ – the GHF n - odder.

On the basis of the model signal matched filter with a transfer coefficient was constructed [5]:

$$\dot{K}_n(\omega,a)=\sum_{n=0}^{140} j^n\sqrt{\frac{\sqrt{\pi}}{n!2^{n-1}}}\exp\left(-0{,}5a^2\omega^2\right)H_n(a\omega)\cdot$$

By analyzing the cross-correlation function between the signal and the model signal, this filter can be used to search for samples with form similar to the model signal in record at Fig.1. Fig.3 shows signal at the filter output that corresponds to the segment recording in Fig.1

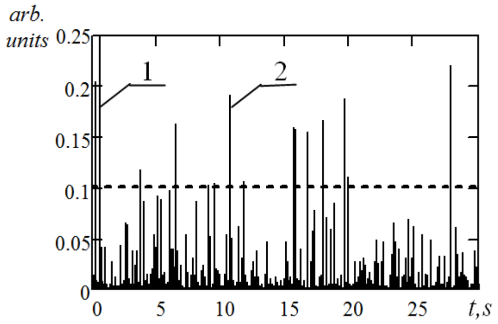


Fig. 3: Cross-correlation function of the signal and model signal

One can single out about a dozen peaks, the level of which exceeds threshold value of 0.1. Peak No. 1 corresponds to the sample, which was used for forming the model signal. Peak No. 2 corresponds to the time ≈10,77 c. The signal in the vicinity of this point is shown in Fig. 4.

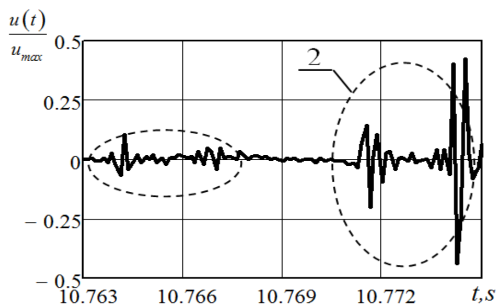


Fig. 4: Sample of the signal in the vicinity of point 2

At fig.4 one can find two fragments, similar to the shape of the model signal. It can be viewed not only as a form of intimacy, but also as a sign of the existence of this fragment of the real term is very close to the model signal.

Hardware and software timing at which the signal sample are closest in form to the model signal can be identified by using the threshold device.

Thus, using the obtained cross-correlation function it is possible to identify the exact location of the selected sample, as well as to identify the cycles of repetition of the signals similar in form to the model signal all over the record.

The best results can be achieved using some criterion of the signal and model signal proximity. As a criterion example, the Cauchy–Schwarz inequality can be used:

$$\left(\int_{-\infty}^{\infty} S(\tau)^2 d\tau\right)^{1/2} \geq \left|\int_{-\infty}^{\infty} S(\tau)W(t-\tau,a)\,d\tau\right|\cdot$$

Minimizing this inequality, we can find the position of the signal on the plane « time t – scale a ».

III. Conclusions

The main advantage of the proposed method is the possibility to design reference signals with almost any shape allowing improving a total diagnosis accuracy. Additionally when parent function is constructed based on the FGH, information about the sample spectral composition in both circular frequency field and in the FGH can be obtained. This feature provides an expert with more data about the process.

Acknowledgment

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Study of Sound Generation in theAarea of Dynamic Trachea Collapse as a Source of Forced Expiratory Tracheal Noise

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Abstract— **Study of sound generation in the area of dynamic trachea collapse is based on: 1) mathematical model of forced expiration mechanics; 2) experimental data on noise generated in the physical model of dynamic trachea narrowing; 3) experimental data on forced expiratory tracheal noise and flow. Flow separation and noise generation in the trachea divergence following the trachea dynamic constriction are considered as a primary source of forced expiratory tracheal noise. Noise generation in the areas of glottis and vortexes arising at airways junction upstream of trachea may substantially contribute to the total upstream of trachea may substantially contribute to the forced expiratory tracheal noise.**

Keywords — **Lung sounds, forced expiration, simulation, trachea, respiratory mechanics.**

I. Introduction

Study of forced expiratory tracheal noise was established for lung evaluation [1]. It was even proved as a better diagnostic tool to discriminate some respiratory disorders as compared with routine flow-volume analysis of forced expiration (FE) [2]. But mechanisms and areas of forced expiratory tracheal noise generation remain vague. Main earlier supposed sources are: 1) turbulent flow noise, 2) vortexes movement downstream of bronchi bifurcations, 3) 3) flow separation and vortexes in the region of extrathoracic trachea expansion following intrathoracic narrowing [1-5].

For analysis of sound generation in the area of dynamic trachea collapse we used: 1) mathematical model of forced expiration mechanics [4]; 2) experimental data on noise generated in the physical model of dynamic trachea narrowing [4]; 3) experimental data on forced expiratory tracheal noise and flow obtained in one volunteer [6].

The purpose of the paper – to study how well the concept of sound generation in the area of dynamic trachea collapse corresponds to experimental data on forced expiratory tracheal noise.

II. Methods

A. Mathematical model of forced expiration

A compartmental model of respiratory system was expanded to simulate sound generation [4]. A pleural pressure is considered as function of time according to the selected maneuver function during forced expiration simulating expiratory effort.

Trachea is simulated as a collapsible tube with nonlinear elastance and linear viscosity.

Pressure difference between pleural and mouth pressures is a sum of pressure differences on viscoelastic lung wall and tracheal segments: upstream and downstream of maximal narrowing. When mechanical parameters of lung and trachea, FE maneuver function are defined, the model of forced expiration mechanics simulates airflow and trachea narrowing upon time during FE.

B. Physical model of trachea narrowing and noise registration in physical model

Physical model of trachea is a rigid tube with diameter 1.8 cm. A diaphragm with centered circular opening could be installed inside the tube. The diaphragm simulates a zone of forced expiratory trachea narrowing. Smoothly with internal tube surface at a distance of 2 cm from diaphragm there was an electrets microphone with frequency band of 20–15000 Hz. In experimental study with this physical model we measured an intensity of noise as a function of diaphragm opening area and airflow. Noise intensity was defined as 0.1 s average of absolute value of microphone output. The intensity characterizes mean sound pressure.

Estimation of tracheal noise dynamics during FE was based on physical model noise data and simulated airflow and trachea narrowing.

C. Data on human FE airflow and tracheal noise

For comparison of estimated and experimental FE airflow and tracheal noise we used experimental data obtained with one volunteer in hyperbaric experiment described in [6]. As soon as above mentioned experiments with physical model were performed with air under normal pressure, only data with the laboratory environment were retrieved form experiment [6]. Simultaneously we registered gas flow by the flow meter ETON-01-22 (Russia) and forced expiratory tracheal noises by virtue of electret microphone with stethoscope nozzle. The volunteer fixed microphone by his hand on the anterior lateral larynx wall. Microphone signal was processed in the same manner as a microphone signal in physical model experiment. Noise intensity upon time was obtained.

III. Results and discussion

Choice of parameters for simulation.

Parameters of the model were chosen in two ways. Lung volumes and airway resistance were measured by total plethismography. Maximal expiratory pressure upon lung volume was obtained by a custom made technique. By adjusting experimental and simulated curves lung volume-time we obtained lung and trachea parameters, i.e. relationship between trachea lumen and transmural pressure, lung elastic recoil, FE maneuver function [4]. Dependence of maximal expiratory pressure upon lung volume was an additional restriction overlaid on FE maneuver function: alveolar pressure during FE is equal or less than alveolar pressure during maximal static expiratory effort.

With fixed lung and trachea parameters and adjusting estimated flow-time data to experimental ones for two other FE maneuvers of the same test subject, we obtained two other FE maneuver functions and dynamics of narrowing in trachea [5]. The minimal ratio of collapsed to initial trachea lumens α, i.e. maximal narrowing was about 0.14

Matching of experimental and simulated dynamics of FE noise intensity.

In order to compare the experimental and simulated dynamics, we adjusted scales so that the simulated and experimental data to coincide at one point in one FE maneuver ($t = 0.3$ s was chosen arbitrary). In all other parts of this and other maneuvers the scales were fixed.

Comparison of simulated and experimental dynamics of FE tracheal sound reveals reasonable qualitative conformity of a decrease of tracheal noise intensity during FE. Discrepancies between experimental and simulated noise dynamics still exist in some part of FE maneuvers. The largest discrepancy was 50% of maximal FE tracheal noise intensity and experimental surpassed simulated twice. In the last part of FE, when simulation predicts no dynamic narrowing of trachea and now tracheal noise, the experiment noise intensity could remain up to 20% of maximal.

Supposed nature of discrepancy of experimental and simulated tracheal noise dynamics

The FE dynamic trachea narrowing is not the only narrowing in the respiratory channel. In the glottis area α~0.2. The narrowing in glottis is almost as significant as in collapsed trachea. Effect of α on noise intensity with $\alpha \approx 0.1$-0.3 was significant in [7] and small in [4]. Anyway one may suggest that sound generated in glottis would be about noise generated in dynamic trachea constriction. As soon as during FE in glottis α does change much, according to [7] we may suggest that glottis noise intensity would be proportional to airflow in the power 2. This component of noise would remain when dynamic constriction and its sound vanishes. Qualitative dynamics of glottis noise upon time would be well comparable with experimental dynamics. Quantitative estimation of contributions of glottis and dynamic trachea constriction was not possible due to absence of sufficient data on absolute levels of noise.

In a real trachea the entrance flow would contain vortexes because of upstream bronchi bifurcations. In the physical model of trachea a special arrangement removed such vortexes. The incoming vortexes may give some contribution to the FE tracheal noise.

IV. Conclusions

Flow separation and noise generation in a trachea divergence following by area of dynamic constriction could be the main mechanism of noise generation during most part of FE. Vortexes in the area of glottis may provide a significant contribution into intensity of FE trachea noise, comparable with the first mechanism. Vortexes born in upstream bronchi bifurcations could contribute to tracheal noise as well.

Acknowledgment

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On the Nature of Elastic Waves Generated by Chest Wall Percussion

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Abstract— **Apparatus chest wall percussion taping and registration of chest surface acceleration was applied to a group of young male volunteers. Velocity of surface waves was determined by a phase technique. In about one half of tests in limited frequency bands within 30 - 350 Hz a phase gradient method for estimation of traveling wave velocity could be applied. Estimated for a few frequency bands wave velocity increased linear with frequency for each test subject. Applicability of phase gradient method could be restricted by generation of additional modes of surface waves.**

Keywords— **percussion, waves velocity, chest wall, phase gradient method**

I. Introduction

Chest percussion is a long known method for medical diagnostics but it is poorly studied from the viewpoint of modern science [1].

In our equipment for automated percussion the surface waves were generated by the computer-controlled indenter [2, 3]. Comparing with our previous works, in the presented paper the equipment capabilities were enhanced, in particular the amplitude frequency characteristics of sensors were improved.

Due to the percussion tapping the vibration of the chest wall appears as well as the surface waves spreading from the point of strike [2, 3]. The objective of this work was to study the characteristics of surface elastic waves propagation in the chest wall of normal subjects.

II. Methods

The experimental setup developed for a study of low-frequency surface elastic waves propagation was described earlier [3]. Briefly, the setup includes a percussion vibrator with accelerometer, the second accelerometer serving as the receiver, and the control block. The vibrator moves back and forth providing percussion taps. The signals from accelerometers were recorded into audio files for analysis.

Twelve male healthy volunteers 19 - 22 years old participated in this study. Measurements were performed on the breathhold at functional residual capacity (FRC).

The vibrations were applied to the skin of the chest wall by the indenter tip. The indenter was glued with double-sided adhesive tape on the right side of the chest at the distance 15-20 cm from the armpit and was slightly hold by hand.

The second accelerometer was glued to the chest wall at a distance from 1.5 to 8 cm from the vibrator. Each measurement lasted 15 s and was repeated 5 times for each subject. The background noise was also measured as the signal from the receiver when vibrations were not generated.

The wave speed was estimated by the phase gradient method [4]. A correlation between two signals was estimated and analysis was performed only for highly correlated signals (correlation coefficient not less than 0.8). The phase shift between signals was determined and the areas with constant slope were selected.

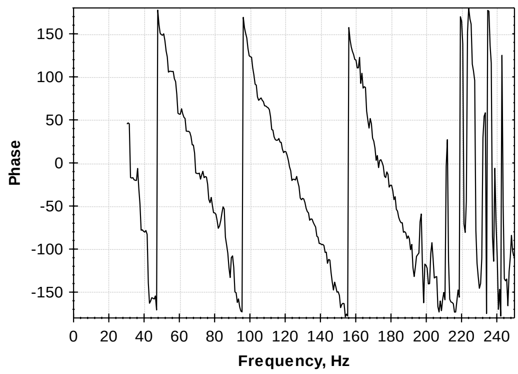

Fig. 1: A representative phase shift between signals (subj. Sh, distance between sensors d=7.25 cm). Two areas with constant slope (50-72 Hz, 110-150 Hz) were selected for velocity calculation

As the slope of phase curve corresponds to the time delay between signals, the wave speed can be calculated for these frequency areas using the distance between sensors. For calculations we used values of distance between the centers of accelerometers which vary from 3.75 to 9.25 cm.

III. Results

The background noise was 20 dB less than the useful signal from the receiver up to 1 kHz in most cases. Two subjects of 12 were excluded from data analysis because of poor coherence between two signals. The coherence between the signals was high in the frequency range 30-200 Hz at least in one study in 10 subjects and at least one area of linear slope of phase vs. frequency was obtained. In 7 subjects there were 10 and more areas with linear relationship between phase shift and frequency providing at least 10 values of wave velocity. Fig. 2 presents a representative example of individual velocity-frequency relationship.

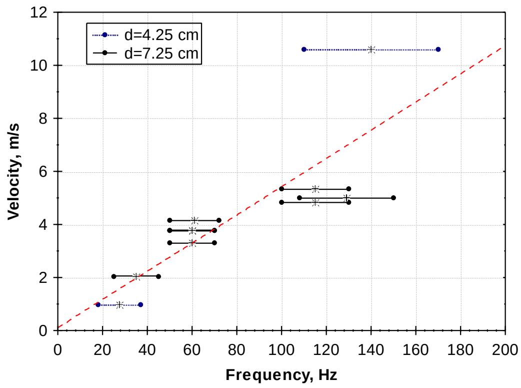

Fig. 2: Individual velocity-frequency relationship (subj. Sh). Areas with linear relationship between phase shift and frequency are depicted by horizontal bars. A center of the bar (depicted) gives one value in velocity-frequency relationship. Linear regression of wave velocity (V) on frequency (F) V=a+b•F with coefficients a= 0.08 m/s, b= 0.053 m/s/Hz, R^2 = 0.76 is presented by dashed line. D – distance between sensors

For all 10 subjects with accepted data a total of 161 velocity values were obtained (Fig. 3). Average individual wave speed rises from 1-5 m/s in low frequencies (30-100 Hz) up to 16-18 m/s in 300-350 Hz. The linear relationship between velocity and frequency results in constant wavelength 4.56±1.42 cm (mean ± SD).

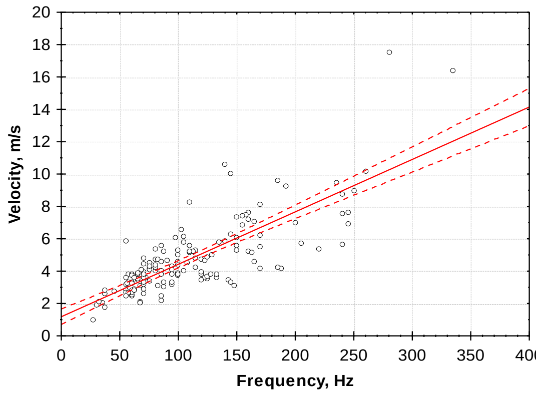

Fig. 3. Velocity of the traveling wave on the chest wall of 10 subjects. Linear regression V=a+b•F, a= 1.18 m/s, b= 0.032 m/s/Hz, R^2 = 0.63 with 95% confidence interval is presented

IV. Discussion

Percussion oscillations of the surface of chest wall propagate in wave-like phenomena, that was proved by cross-correlation study of accelerations of intender and receiver [2]. A pilot study with the phase gradient method revealed wave velocity increase with frequency [3].

In this study with a larger group of test subjects we estimated applicability of phase gradient method to waves on the surface of the chest wall.

If just one type (mode) of wave with frequency-independent speed propagates in media, then phase-frequency relationship would be linear like in denoted areas in Fig. 1. But this relationship was only in about one half of tests, in limited frequency bands.

Percussion waves propagates in a multilayered structure of soft tissue, rib cage and pulmonary parenchyma. Theoretical studies of surface waves propagation in layered structures [4, 5] revealed a complex nature of propagation. A few modes of waves appear with increasing frequency. In a viscoelastic layer of thickness H and tranverse wave velocity Ct, attached to a rigid basement, the crucial parameter is ωH/Ct, where ω is angular frequency [4]. If 1.5 < ωH/Ct < 4, there is only one mode, no wave if 1.5 > ωH/Ct and a few modes if ωH/Ct > 4. With H=0.5 cm, Ct=2.5 m/s, the second mode would appear at the frequency about 300 Hz with more modes at higher frequencies. We suggest that percussion generates one-mode traveling wave in a limited range of frequencies. Waves velocity increase could be in our experiment because of a higher mode component or effect of tissue viscosity with the same mode.

V. Conclusions

In about one half of tests in limited frequency bands within 30 - 350 Hz a phase method for estimation of traveling wave velocity could be applied. Estimated for a few frequency bands wave velocity increased linearly with frequency for each test subject. Applicability of phase method could be restricted by generation of additional modes of surface waves.

Acknowledgment

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Measuring CABS in Obstructive Airways Diseases

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Abstract— **Respiratory Sounds (RS) auscultation is the main tool for ongoing lung condition evaluation. While normal RS are characterized as broad-band sounds, the abnormal RS would usually include some form of adventitious sounds continuous or non continuous. In this review the nomenclature, clinical context and methods for detection of Continuous Adventitious Breath Sounds (CABS) are described.**

Keywords— **Continuous, Adventitious, Breath, Sounds**

The term Continuous Adventitious Breath Sounds (CABS) was coined in 1987 by members of the International Lung Sounds Association to identify a group of breath sounds that are (a) superimposed on the basic breath sounds; and (b) continuous. Whereas the basic breath sounds are characterized as broad-band sounds, the CABS are narrow band in the frequency domain. Their sonogram typically contains ridges that are narrow and sharp and last >80 msec. Table 1 shows the known kinds of CABS and their spectral characteristics:

Table 1: Summary Characteristics of CABS

Name	**Description**	**Clinical Context**	**Comments**
Wheeze	Inspiratory or expiratory, frequency range 150 to 1500 Hz, may be multiple coinciding wheezes. May have a harmonic or two. Best heard over the trachea	Associated with airway obstruction. If relieved by a bronchodilator, is likely to be due to asthma. Other conditions include COPD, foreign body etc.	Except in small infants, not audible in the environment. Associated with regional airway flow limitation
Forced expiratory Wheeze	Expiratory. Frequency range 150-1500 Hz. Best heard over the trachea	Vocal Cord Dysfunction, not relieved by bronchodilator.	Audible in the environment. Associated with global flow limitation
Rhonchus	Inspiratory or expiratory, frequency range 80 to 250 Hz, multiple (i.e. >3) harmonics.	Associated with fixed airway obstruction typically with higher compliance airways (collapsible), e.g. bronchiectasis.	May be cleared by cough
Whistle	Inspiratory or expiratory, frequency range >1200 Hz with rapid frequency changes with flowrate. Only heard over the trachea and at the mouth.	Associated with airway obstruction. If relived by a bronchodilator, is likely to be due to asthma. Other conditions include COPD.	Whistles cannot be heard by auscultation with a stethoscope over the chest wall.
Stridor	Inspiratory or expiratory, frequency range 150 to 1500 Hz, heard best over the trachea, audible in the environment.	Associated with upper airway obstruction (e.g. vocal box edema, epiglottitis). Relieved with racemic epi and steroids.	Often associated with concomitant bronchoconstriction and wheeze.
Snore	Inspiratory or expiratory, frequency range 80 to 250 Hz, multiple harmonics, audible in the environment.	Associated with dynamic collapse of the pharyngeal upper airway causing flutter of the walls of the air-channel	Eliminated with CPAP

We shall describe the methods used to detect CABS and the criterial for sorting them out, including the methods used to differentiate between those CABS who are confined to the chest and those who are also present in the environment. We shall also outline the methodology used to validate CABS detection.

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Quantitative CABS Monitoring in Asthma Management

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Abstract— **Asthma is a chronic disease characterized by attacks of reversible narrowing of the lung airways. Traditionally, repeated measurements of peak expiratory flow (PEF) have been advocated as a tool for asthma self-management. However, the shortcomings of the PEF include the need for full cooperation and perfect technique. The feasibility of using quantitative Continuous Adventitious Breath Sounds monitoring, as an equivalent tool for monitoring and managing asthma is reviewed. This tool does not require patient cooperation.**

Keywords— **Asthma, Wheeze, Acoustics, Monitoring**

Asthma is a chronic disease characterized by attacks of reversible narrowing of the lung airways that can be severe or even life-threatening at times. While essentially incurable, asthma can be managed and controlled using broncho-dilators and anti-inflammatory medications. The key to modern asthma management is to give the patient just enough medications to prevent attacks. To do so, there should be simple, objective and inexpensive tools for assessing and monitoring the patient's asthma status, preferably at the patient's own environment. Repeated measurements of peak expiratory flow (PEF) has been advocated as such tool for asthma self-management in the last >20 years within the framework of an Asthma Action Plan (AAP). However, certain shortcomings of PEF measurements are well recognized: it requires full cooperation and perfect technique to be accurate and repeatable, as such, it is not practical in small children, patients with disabilities, the elderly and the weak. The goal of this presentation is to review the feasibility of using quantitative continuous adventitious breath sounds (CABS) monitoring as an equivalent tool for monitoring and managing asthma. The primary working hypothesis is that changes in the WheezeRate™ (Wz% = 100*Tw/Ttot, Baughman & Loudon 1984) measured by standardized tools (e.g. acoustic respiratory monitor (ARM™) algorithm; AirSonea®, iSonea Ltd, Melbourne, Australia) follow the pattern of changes of PEF or FEV1.0 when the degree of airway constriction changes in an asthma patient. Such changes can be a result of administering a rescue medication (bronchodilator) or during a bronchial challenge test using methacholine, exercise or cold air. Additional criteria for confirming the validity of using Wz% in the monitoring and assessment of asthma patients are as follows:

- Normal patients have zero or very low (e.g. <3%) WheezeRate.
- Asthmatic patients who are well controlled have very low (e.g. <3%) WheezeRate during both wakefulness and sleep time.
- Asthmatic patients who are not well-controlled have elevated WheezeRate. This can be during wakefulness and/or during sleep.
- Elevated WheezeRate is always associated with airway obstruction, but only when it is reversible with a bronchodilator it is a specific sign of asthma.
- Asthmatic patients in respiratory distress with little or no wheeze whose WheezeRate increase following administration of a bronchodilator may have a "Silent Lung", a form of severe asthma attack.

Data from clinical studies that document these criteria will be presented.

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Possibilities of the Computer Bronchophonography in Diagnostics of Malfunction of External Respiration at Children with Cystic Fibrosis

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Abstract— **Computer bronchophonography reveals the presence of obstructive disorders in children with cystic fibrosis under 6 years as well as debilitated patients, not capable of carrying out the forced respiratory maneuvers. Our study demonstrated significantly higher levels of acoustic work of breathing in low, medium and high frequency ranges, in children older than 6 years with cystic fibrosis, compared with healthy peers. Children under 6 years with cystic fibrosis are characterized by a strengthening of acoustic work of breathing in the low frequency range. The initial symptoms of bronchial obstruction in this age group reflects significantly higher values of the relative ratios. The presence of bronchiectasis and chronic Pseudomonas aeruginosa infection leads to deterioration of respiratory function due to the significantly higher values of the acoustic work of breathing in the high frequency range.**

Keywords— **Cystic fibrosis, children, function of external respiration, spirography, computer bronchophonography**

I. Introduction

Cystic fibrosis (CF) is the most frequent hereditary autosomal-recessive disease at persons of Caucasian race. In 2011 in the Russian Federation the average frequency of a disease was 1:10498 that is significantly lower, than in European countries [1]. In the Siberian Federal District, according to neonatal screening of 2006-2012, general frequency of CF over region makes 1:7467, in Omsk region in particular - 1:8594 [2, 3].

Chronic pulmonary involvement is the most serious clinical implications of CF. Changes of bronchopulmonary system, emerging in the first weeks of a child's life, gradually lead to obstruction of small airways [4]. Bronchiectasis plays an important role in the pathogenesis of bronchial obstruction at CF. Thinned walls of bronchi lose their elasticity – on the inhale they are stretched, and on the exhale they deflate, thereby increasing obstruction [1].

In CF children of the first three years of life in a phlegm Staphylococcus aureus is mostly often sowed. With age there is sequential colonization of bronchial tree with gram-negative flora, including Pseudomonas aeruginosa [1].

Measurement of respiratory function by method of spirography is important part of clinical trials and control of extent of development of CF. However need of performance of the meaningful forced respiratory movements makes impossible its applications for small children. This problem may be solved with a help of computer bronchophonography - the method based on registration of a respiratory cycle and the analysis of amplitude-frequency characteristics of respiratory murmurs [4].

The objective is to estimate respiratory function at children with CF, by means of computer bronchophonography.

II. Materials and methods

At the Omsk CF Center by means of computer bronchophonography were examined 34 children aged from 3 months till 17 years, among them 18 boys, 16 girls. The diagnosis of CF in 100% of cases was confirmed by data of sweat, molecular and genetic tests.

The severe course of a disease was characteristic for 22 children (65%), while in 12 (35%) it proceeded in a moderately severe to severe form. Most of the patients were children of school age and teenagers (20 persons, 11,5 [9; 15] years), 14 examined children were under the age of 6 years (2,0 [1; 3] years). During examination microbiological research of phlegm, computer bronchophonography and the spirography (for children older than 6 years), multispiral computer tomography were performed.

Control group of 42 children aged from 4 months till 17 years, without any indications in the anamnesis of acute or chronic respiratory diseases, non-smokers, included 22 girls and 20 boys. The examination was performed in 16 children under the age of 6 years (2,0 [1; 3.5] years), 26 children of school age and teenagers (11,0 [8; 14] years). Respiratory function in control group was estimated by means of computer bronchophonography.

For children about one year recording respiratory murmurs of quiet breathing was made in a prone position, while in older children and adults it was made in a sitting position. Respiration registration at children under 2 years was conducted by means of facepiece, which was connected to the sensor unit, at patients older than 2 years a biteboard was attached to the sensor, a clip was imposed on a nose. Record was carried out for 10 sec. No less than three attempts were applied to obtain reproducible results.

During computer bronchophonography time series curve is fixed which is proportional to acoustic noise produced during breathing. A quantitative assessment of energy costs of bronchopulmonary system to specific acoustic excitation of the phenomenon throughout the respiratory cycle, or its separate phases - an acoustic work of breathing (AWB) is estimated. Scanning respiratory cycle is carried out in the frequency range of 200 Hz to 12600 Hz. In order to avoid the masking effect of low frequency cardiac noise (up to 200 Hz) in the set features a special cut-off low-pass filter is used. There are three areas of the frequency spectrum: 200-1200 Hz (low-frequency range, AWB1), >1200-5000 Hz (mid-range AWB3), >5000-12600 Hz (high frequency range AWB2). Area of low frequencies (up to 1200 Hz) describes the state of the upper respiratory tract. The zone of high frequencies (above 5000 Hz) reflects the obstructive changes in the lower respiratory tract, and is correlated with the sound characteristics of the wheezes [4]. K - coefficient, which reflects the same parameters in relative terms: the spectrum of low frequencies - K1 = (AWB2 + AWB3)/AWB1 x100; high range - K2 = AWB2/AWB1 x100; midrange - K3 = AWB3/AWB1 x100 [4].

III. Results and discussion

The main group and control group were comparable on age and a gender (Mann-Whitney, $p>0,05$). In control group statistically significant difference in indicators of AWB depending on age was not revealed, therefore the total result was estimated.

Statistically significant higher AWB at children with CF in high-frequency range in comparison with healthy ones is noted (Kruskal — Wallis, $p\leq 0,05$). This indicates on a preferential distal respiratory tract at children with CF even without exacerbation of a disease due to the phenomena of mucostasis, deformation and the fibrotic changes of walls of the bronchi.

In comparison with healthy coevals, more expressed changes of respiratory function are registered at children older than 6 years. Statistically significant higher AWB is noted in all frequency ranges (AWB1, AWB3, AWB2) that reflects total involvement of structures of tracheobronchial tree in pathological process (Mann-Whitney, $p<0,05$). The severity of bronchial obstruction emphasize significantly higher values of the relative coefficient K2 (Mann-Whitney, $p<0.05$).

At children of age under 6 years, patients with CF, disorders of respiratory function in comparison with healthy coevals are less expressed. At children of this age significantly higher AWB at low-frequencies range that indicates an affection of upper airways is registered (Mann-Whitney, $p<0.05$). It is caused by gradual formation of irreversible changes of tracheobronchial tree at children due to chronic inflammation. Statistically more significant relative values of coefficients of K1, K2, K3 confirm the conclusion (Mann-Whitney, $p<0.05$).

At 16 children older then 6 years (80%) the severe course of CF was noted. The examination revealed significantly lower values of forced expiratory flow in 1 second in % of norm (FEV1%) in comparison with moderately severe patients. In this age group at severe course of CF formation of bronchiectasis occures significantly more often (Mann-Whitney, $p<0.01$).

The presence of bronchiectasis is accompanied by statistically significant increase of AWB in all frequency ranges, especially high frequency, decrease of FEV1%, the development of chronic Pseudomonas aeruginosa infection (Mann-Whitney, $p<0.01$).

For children of age under 6 years severe course of a disease is less characteristic - 43% (6 children). It isn't revealed statistically significant differences in microflora of airways, frequency of development bronchiolo - and bronchiectasis depending on severity of a disease. Severe course in this age group is characterized by significantly higher AWB values in the mid-frequency range, that reflects involvement in pathological process mainly bronchi tubes of medium caliber. Relative coefficients incresed.

Regardless of sex and age of a child, for severe course of CF, constant presence of auscultatory symptoms involving mixed wet rales (Mann-Whitney, $p<0.01$) is found. Respiratory tract of patients is significantly more frequently (Mann-Whitney, $p<0.01$) affected by chronic colonization by Pseudomonas aeruginosa.

IV. Conclusions

The expressed changes of respiratory function according to computer bronchophonography are found for all children with CF, progressing with age. Lack of damage to the respiratory tract in the form of bronchiectasis and chronic Pseudomonas aeruginosa infection determines higher rates of respiratory function. Computer bronchophonography can be recommended as a method of diagnostics of disorders of respiratory function at children of age under 6 years, who are patients with CF.

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Computer Bronchophonography as a New Method for Lung Function Assessment in Children with Bronchopulmonary Diseases

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Abstract—**Background: Spirometry is a widely used method for evaluation of lung function in 7-year-old children and older with obstructive or restrictive lung diseases, such as asthma or cystic fibrosis. However, it is still necessary to develop methods for assessment of respiratory function in younger children.**

Aim: To introduce computer bronchophonography (CBPhG) as a method for assessment of respiratory function.

Materials and methods: One hundred forty four healthy subjects (from 1 to 15 years old) without lung diseases underwent CBPhG. Respiratory sounds were recorded with a highly sensitive transducer in a wide range of frequencies (low 0.2–1.2, middle 1.2–5.0, and high frequencies > 5.0 kHz) in normal breathing. The registration was performed for 10 sec.

Results: Obtained reference values of acoustic componentin children without lung diseases, regardless of age were 0.24±0.08mJ in a high frequency range, 3.04±0.13mJ - middle, 71.7±9.8mJ – low.The CBPhG showed sensitivity 86.4 %, and specificity 90.9 %.

Conclusion: This method can be used for further functional and clinical diagnostics of chronic and acute broncho-pulmonary diseases in children.

Keywords— **asthma, children, lung sounds, pulmonary function assessment, computer bronchophonography, pattern.**

I. Introduction

It is quite common for children to suffer from various acute, recurrent, and chronic respiratory diseases during the first 5 years of their life. Auscultative data and the examination of respiratory function play a significant role in diagnostics and treatment monitoring of bronchopulmonary diseases [1,2]. About 60–80 % of children experience manifestation of asthma within the first five years of life. Pulmonary diseases are characterized by irregular breathing patterns and presence of abnormal pulmonary sounds such as wheezes and crackles. Air flow in respiratory tract has a high speed and is turbulent in nature. This leads to high and low frequency acoustic phenomena which can be detected by acoustic scanning of breath [3,4]. Wheezing is formed due to increasing turbulence in the airways which are enhanced by bronchoconstriction, structural changes of epithelium, edema, and sputum. In order to detect abnormalities in the respiratory system it is important to define reference indicators of respiratory pattern in children of all ages.

II. Materials and methods

A. Methods

The computer bronchophonography (CBPhG) has been developed by Malyshev V.S. et al. and adapted for pediatric respiratory practice by Kaganov S.J., Geppe N.A., et al. This is a method for evaluation of breathing pattern in children, when it is hard to do functional lung tests [5,6]. The CBPhG can be performed in children of all ages, because it does not require respiratory maneuvers.

The CBPhG is based on a spectral analysis of respiratory sounds at mouth. It allows to assess a breathing pattern and recognize respiratory sounds by recording (scanning) different acoustic features of a respiratory cycle. The mathematical analysis was performed with a special software based on Fast Fourier Transform. The CBPhG comprises the analysis of time and frequency of respiratory sounds spectrum which relates to changes of airflow in the bronchi.

This method allows recording a time curve of the sound produced in the respiratory tract due to air turbulence. We named it an acoustic component of the work of breathing (AC) (measured in mJ). It helps to evaluate the sound characteristics of the respiratory rate (spectral density).

Recording the acoustic phenomena, which occur due to changes in the airflow turbulence in breathing, is carried out by high sensitive sensor. This sensor is embedded in a special mouthpiece, which a baby holds in his/her mouth. In order to eliminate cardiac sounds the measuring system has a restrictive filter with cut-off frequency of 200 Hz. Moreover, the system contains two sets of filters. The first one provides a sample of signals in a frequency range of 200–5000 Hz, whereas the second one –in a range of 5000–12000 Hz. The signal from the microphone reaches a preamplifier which amplifies it up to 0.8 V; the output signal is then subsequently processed.

During the investigation the subject should breath normally, without any maneuvers, but not scream or cry. The procedure can be performed even during the first months of child's life. Registration takes about 10 sec. Using the headfhones we recorded the baby's breath at the beginning of his exhalation. The assessment was repeated three times in order to exclude any artifacts in the curve. Afterwards, the respiratory cycle was visually screened for significant peaks in different parts of the frequency range, and then we selected the part with pronounced oscillations during 4 sec. This graphical representation is known as "Respiratory Pattern" (Fig.1).

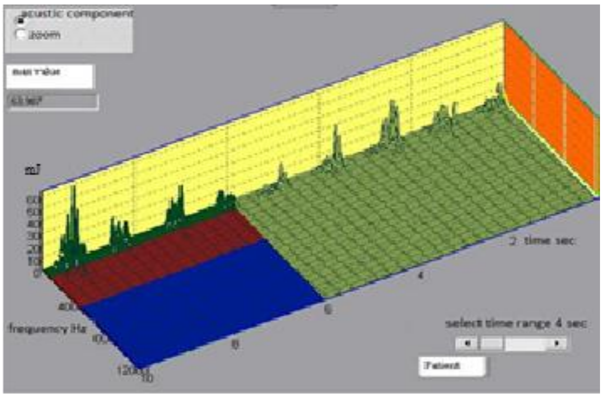

Fig. 1: Ten-seconds bronchophonogram in the three-dimensional view

B. Materials

We assessed the lung function in children of different ages (ranging from 1 to 15 years old) using the CBPhG. The examined children had no abnormalities of the respiratory system and the history of recurrent and chronic respiratory, cardiovascular diseases as well as diseases of ears, nose and throat. They did not experience any respiratory disease 2 months before the examination. Children older than 6 years had results compared with the data of spirography (FVC, FEV1, FVC/FEV1, PEF). Bronchophonography was performed during normal breathing prior to spirography, because the latter includes a forced expiratory maneuver and may alter the sound phenomena of respiration.

For every child four bronchophonograms were recorded: baseline, at 10th minute, 1 hour, and 1 week later. Reproducibility of acquired results was 97 %. Artificial neural network with ROC-analysis was applied to assess the diagnostic efficiency of CBPhG [7]. Pre-built artificial neural network was used to evaluate sensitivity and specificity of the method [3].

We examined 144 children: 52 children from 1 to 5 years old, 48 children 6–9 years old, and 44 children 10–15 years old (55 % boys and 45 % girls). The results of the routine lung function tests in children older than 6 were within normal range.

III. Results

Using the CBPhG we analyzed a full spectrum of frequencies within range of 0.2–12.6 kHz: low 0.2-1.2, middle 1.2-5.0, and high frequency > 5.0 kHz. It was found, that children in different frequency ranges have a meaningful estimate of the acoustic component of the work of breathing (AC).

In the respiratory cycle of healthy children oscillations of the sound is much higher in the low frequency range, than in the mid frequencies. Regardless of age in bronchophonograms we did not detect any significant oscillation in the high frequency range(> 5 kHz); average AC was 0.24 +/–0.08 mJ. However, high frequency range had diagnostic value in children with airway obstruction. Identification of sounds in the high frequency part of the spectrum suggests a presence of constrained diffuse bronchial patency, resulting in a substantial change in air-flow of exhaled air [8].The CBPhG showed sensitivity 86.4 %, and specificity 90.9 %.

Variations in the middle range frequencies reflect the acoustic phenomena of the bronchi, trachea, and nasopharynx. Reference indicators were 3.04 +/- 0.13 mJ and 71.7 +/- 9.8mJ in the middle and low frequency ranges, respectively. According to the results of our study pathologies of the upper respiratory tract show that the oscillation in low frequency range AC (0.2-1.2 kHz) increase [9]. The pathology of the lower respiratory tract is rarely isolated, so increased oscillations are recorded in all frequency ranges.

IV. Conclusion

Respiratory soundcharacteristics, obtained by computed diagnostic device, can serve as an integral individualized evaluation parameter (pattern). This method can be used for further functional and clinical diagnostics of chronic and acute broncho-pulmonary diseases in children.

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Comparison of Inspiratory Chest Lung Sounds in Patients with Asthma, COPD and Healthy lungs

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Abstract— **In the research inspiratory chest lung sounds in patients with asthma, COPD and healthy lungs were compared using single-point parameters of power spectral density (frequency limits F_{50} (median frequency), F_{75}, F_{90}, lung sound amplitude in terms of root mean square (RMS) value of the total spectral power) and duration of inspiration. It was observed that the value of RMS parameter in patients with COPD is lower than in both asthma and control group. The inspiration was longer in COPD group, the shortest duration was observed within control group.**

Keywords— **COPD, asthma, spectral analysis, lung sound, inspiration.**

I. Introduction

Asthma and COPD have their own risk factors, pathogenetic mechanisms, diagnostic criteria, but in some cases it remains difficult to differentiate between chronic obstructive pulmonary disease (COPD) and asthma in clinical practice, especially in a primary care settings [1]. The idea that computer analysis of breath sounds can introduce new significant criteria for the improvement of differential diagnosis has been considered in several researches before. Thus Malmberg et all. [2] reported that in in flow standardized inspiratory breath sounds the median frequency of the spectra and the total spectral power in terms of root mean square were higher in asthmatics than in both control patients and patients with COPD. Guseinov et all. observed the differences between patients with COPD and asthma in the value of area under the spectrogram curve in high-frequency range (5 – 12,6 kHz) [3]. The aim of the current study was to assess if the using of the most market available and well-known between clinicians electronic stethoscope Littmann 3200 with following spectral analysis of the signal (the scheme that can be easily implemented in routine medical practice) can reveal additional diagnostic criteria for the issued problem.

II. Methods

The following groups of patients were considered (the diagnosis was confirmed clinically): 1. COPD (mild to moderate stage, age 48 to 69 years, mean age 57; active and former smokers); 2. Asthma (mild, medium, severe stage; age 59 to 78; mean age 67.1; non-smoking); 3. The control group: people without disease of the respiratory tract (non-smoking; age 29 to 71; mean age 40.2). Recording was done before taking of bronchodilators, all patients were in remission stages. Breath sounds were recorded in sitting posittion in four symmetrical points: both from left and right sides at the level of the 2nd rib in the chest and at the level of the 7th rib near blade angle in the back. The Littmann 3200 stethoscope in extended filter mode (amplifies sounds from 20 – 2000 Hz, emphasizes the sounds between 50 – 500 Hz) was used, additionally the signal was digitally filtered in the frequency range 100 – 1500 Hz. Patients were asked to breathe quiet, deep, not causing cough or shortness of breath to provide relatively standard experiment conditions. The length of records was 18 seconds.

Unfortunately not all recorded sounds were taken into the analysis. The first reason for the exclusion was considerable heart sounds during all recording length. Also records or part of records with lots of artifacts (caused by friction of stethoscope chestpiece or cable on a patient body) were deleted. The expiratory sounds were not considered because breathing pattern strongly depends on the extent of bronchial obstruction [2] and not in all signals the beginning and the end of exhalation could be determined clearly using only lung sound channel. Thus the quantity of the patient in the considered groups was the following: COPD – 8 patients, asthma – 11 patients, control group – 13 patients. Inspiratory phases were manually extracted using sound and visual analysis of spectrogram, the formal criteria for the detection of starting and ending points was determined by the stethoscope inherent noise (-40 dB relatively to the maximum spectral count).

Inspiratory phases from 3 – 8 breathing cycles were averaged, spectral analysis of the average signal using Welch method was performed (512-point FFT, 50% overlapping of adjacent Hanning data windows). To describe the acquired power spectral density (PSD) following single-point parameters were used: frequency limits (F_{50} – median frequency; F_{75}; F_{90}), lung sound amplitude in terms of root mean square (RMS, $X_{rms} = \sqrt{\frac{1}{N}\sum_{n=1}^{N}|X_n|^2}$, where X is the amplitude of PSD samples) value of the total spectral power (the area under the curve in a certain sense). The choice of these criteria was determined by the experience of the previous research [2]. The average duration of inspiratory of each record was also analyzed. The breath sounds variables between the studied groups were compared by the non-paired Mann-Whitney U-test, a p-value of less than 0.05 was considered significant.

III. Results and discussion

The preliminary analysis of PSD showed that the frequency range of most signals, regardless of patient group and points of record, was concentrated in the range of 50 to 340 Hz. It corresponds well with the fact that Littmann 3200 stethoscope working in extended frequency mode emphasizes the sounds between 50 – 500 Hz.

Table 1: The values and confidence intervals (p = 0,05) of calculated parameters in different studied groups

	COPD	Asthma	Controls
RMS (dB)	2,34 ± 1,47	5,46 ± 2,57	5,34 ± 2,85
F_{50} (Hz)	107,6 ± 9,4	110,1 ± 5,5	106,3 ± 3,9
F_{75} (Hz)	158,7 ± 10,7	141,0 ± 10,6	133,9 ± 10,2
F_{90} (Hz)	319,11 ± 60,2	350,49 ± 25,2	275,6 ± 26,3
Ins. dur. (sec)	1,17 ± 0,06	1,07 ± 0,06	0,85 ± 0,03

The best results (according U-test) between considered parameters were observed in the RMS value of the total spectral power and the averaged inspiratory duration. As it can be seen in Table 1 and Figure 1 the lowest RMS value was observed in patients with COPD, the highest in patients with asthma and the RMS value of healthy people is close to the value in asthma patients. The result can be explained by the fact that in COPD respiratory tract obstruction is formed at the level of small and very small bronchioles (6th – 8th order of bronchi and bronchioles) and thus is characterized by higher frequency content of lung sounds. Changed lung tissue (parenchyma) suppresses high-frequency components and as the result weak breath is observed in COPD patients. In asthma patients obstruction is mainly occurred in larger airways segments and parenchyma impact is not so significant.

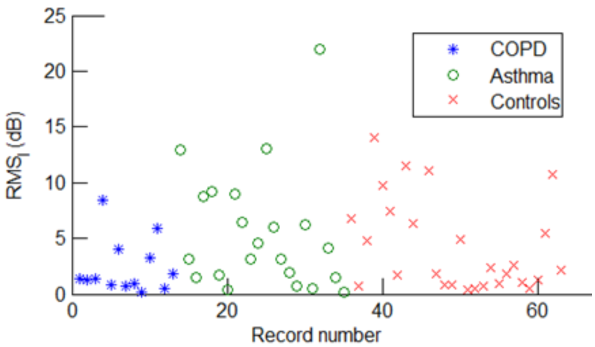

Fig.1: Distribution of inspiratory root mean square value of the total spectral power of breath sounds in patients with COPD, asthma and healthy lungs (controls)

In contrast to the results presented in [1] we have not found differences between the values of median frequencies within studied groups.

It was also observed that the duration of the inspiratory in COPD patients is 9% bigger than in patients with asthma and 27% bigger compared with controls (Table 1, Figure 2). This phenomenon should be studied more on a larger amount of data, because usually COPD patients are characterized by shortness of breath and the increasing of expiration duration. Perhaps the received effect can be explained either by psychological control of breath (during the examination COPD patients might inhale deeper than usual) or by the different stage of bronchial obstruction among the studied patients.

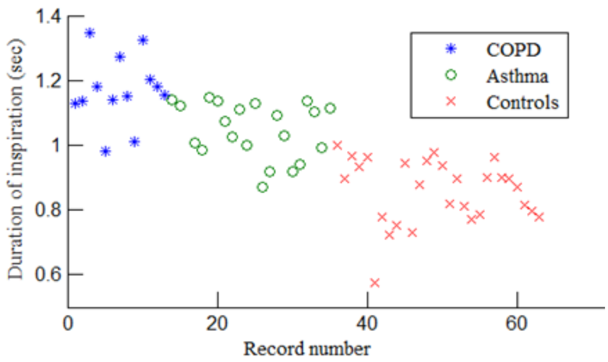

Fig.2: Distribution of inspiration duration in patients with COPD, asthma and healthy lungs (controls)

IV. Conclusion

The following conclusions can be made on the basis of the executed research:

1. It was observed that the intensity of inspiratory chest sounds is significantly lower and the average inspiratory is longer within COPD than in asthma group. This fact points to the possibility of using the RMS as a prospective diagnostic feature for differential diagnosis of the disease, the effect of inspiratory prolongation in COPD patients should be studied more carefully on a larger amount of data.
2. The performance of median frequency as the parameter of COPD and asthma differential diagnosis will be studied in more details during subsequent researches.
3. For the further research it is necessary to use electronic stethoscope with linear frequency response, better noise immunity, less sensitivity to heart sounds. Also the synchronous registration of inspiration and expiration flow will standardize experiment condition, automate and increase signal processing performance.

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Computerized Respiratory Sounds Analysis on the Basis of Multilevel Wavelet Transform

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Abstract— **Software “WaveView MWA” has been developed to build wavelet sonograms with a high frequency-time resolution. Records of the textbook “Auscultation of the Lungs” have been processed using a high-precision algorithm for multilevel wavelet analysis. Examples of wavelet sonograms, frequency-time characteristics of lung, respiratory sounds and forced expiration sounds have been obtained. The multilevel wavelet transform based sonograms are shown to provide a higher frequency-time resolution than Fourier spectrograms.**

Keywords— **computerized respiratory sounds analysis, multilevel wavelet transform, Fourier spectrogram, wavelet sonogram**

I. INTRODUCTION

Data obtained by Russian and foreign researchers over the past 10 years show that lung sounds, respiratory sounds and forced expiration sounds should be classified as complex non-stationary signals. It is known that the human acoustic reception does not physically perceive the low-frequency region of lung sounds. An objective study of the acoustic properties of the lungs was made possible after computerized methods of data recording and processing came in practice. Introduction of modern computer analysis technology into the therapeutic clinic made it possible to obtain new information on the characteristic features of lung sounds. A number of devices have been developed for automated diagnosis of respiratory sounds, which is important in the early stages of identification of critical conditions in a subject in pulmonology, acoustic mapping of respiratory sounds, modeling of respiratory sounds, and the study of their source. In this regard, objectifying information obtained by the new methods of computer digital auscultation of human breath sounds is relevant in the biomedical acoustics.

The objective of this study is a comparative analysis of qualitive indicators of frequency-time parameters of lung sounds based on Fourier spectrograms made by computer phono-spirographic systems that have been widely used in recent years, and wavelet sonograms or “visible sound” images computed with new technology of signal multilevel wavelet analysis.

II. MATERIALS AND METHODS

A. Respiratory sounds

The spectral components of lung sounds, respiratory sounds and forced expiration sounds are in the frequency range between 3-5 Hz and 5000 Hz. The duration is from several milliseconds to tens or hundreds of milliseconds [1].

B. Registration of respiratory sounds

Registration of respiratory sounds on the surface of the chest wall. Respiratory acoustical signal pickup (includes a miniature Logitech USB microphone and an analog-digital convertion module). Computer software for signal input (“Registrator 2.11” software ensures input of signals into a personal computer, editing of the registered files, archiving and database maintenance).

II. RESULTS

The Fourier transform is the most commonly used spectral analysis algorithm to provide information on the frequency components of a respiratory sounds [2]. Computer analysis of lung sounds is performed, usually based on Fourier transform, have disadvantages at processing non-stationary signals [3, 4]. Figure 1 shows a Fourier spectrogram of an infant’s breathing sound [5]. It shows a poor frequency-time resolution of lung sounds in the low frequency region.

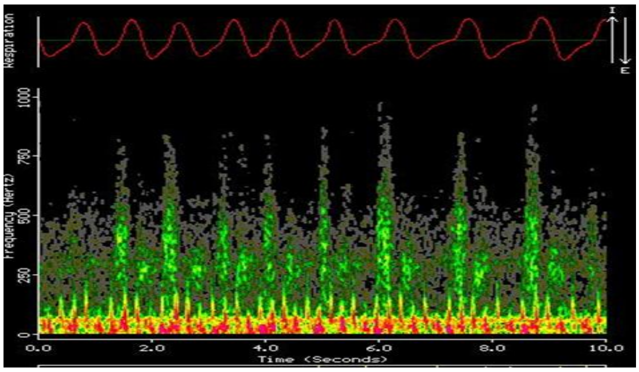

Fig. 1: Fourier spectrogram of an infant’s breathing sound

Multilevel wavelet analysis

Customized software for obtaining wavelet sonograms (“WaveView MWA” [6]) carries out multilevel wavelet-transformation of respiratory sounds according to the set-up parameters, their display and record. A high frequency-time resolution of “visible sound” images is achieved using a signal multilevel wavelet transform. The used mother wavelets include: Morlet, Haar, “Mexican hat”. Testing of the software “WaveView MWA” showed that it was possible to extract non-stationary low-level signals up to –60 dB. The wavelet sonograms obtained by “WaveView

MWA” customized software is shown in Figure 2 and Figure 3.

Figure 2 is a wavelet sonogram of the same infant’s breathing sound [5]. It shows a high frequency-time resolution of lung sounds in the given frequency range. The cardiocycles of the infant’s heart sounds are seen in the low-frequency region.

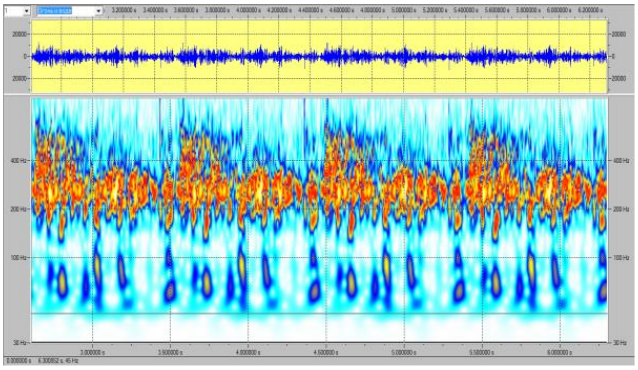

Fig. 2: Wavelet sonogram of the same infant’s breathing sound

Figure 3 is a wavelet sonogram of forced expiration sounds.

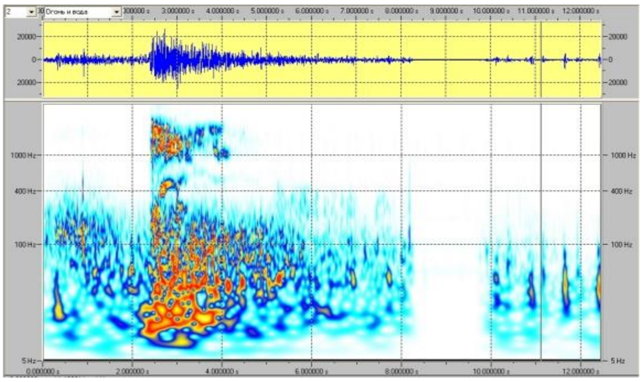

Fig. 3: Wavelet sonogram of forced expiration sounds

The time interval of 0 to 2 seconds characterizes the inspiration. The interval from 2 to 8 seconds characterizes forced expiration. In the interval from 10 to 13 seconds the heart function is demonstrated.

The developed multilevel wavelet transform based software “WaveView MWA” provides a higher frequency-time resolution of lung and respiratory sounds than the Fourier transform [6].

III. Conclusions

The multilevel wavelet analysis of lung acoustic field signals allows obtaining frequency-time descriptions/wavelet sonograms of lung sounds and noises with a far higher resolution than that of spectrograms of phono-spirographic computer systems using the Fourier transform. Wavelet sonograms provide visual objective and complete information in the study of lung and respiratory sounds.

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Small Pneumothorax Managed Without Chest Tube Insertion With Guide of Lung Sound Mapping in 2 Young Patients with Polycystic Lung Disease

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Abstract— **We encountered 2 young Japanese males with spontaneous Pneumothorax; both had underlying Polycystic Lung disease and had experienced episodes of spontaneous pneumothoraxes which were managed by chest tube insertions. One patient actually developed lung-pleural fistula as complication of the insertion of chest tubes into the existing bulla, which required surgical repair, but next few years they were asymptomatic and whatever non-strenuous-athletic activities they did they could tolerate.**

Upon arrival to my office, both patients had symptoms of spontaneous pneumothorax, however both had reasonable air movement throughout lung fields by auscultation, and not hypoxic which was clearly shown on breath sound recordings with a lung mapping by STG-16 Murphy.

X-rays revealed only about 12% to 15% pneumothorax with no sign of tension pneumothorax, therefore it was decided not to insert chest tubes and observe closely with serial lung sounds mapping. In case any changes of symptoms he can get in touch with us by any means of beepers 24 hours a day. In both cases the course was satisfactory and they were able to return to school with no new pneumothorax with the following few years of follow up.

Keywords— **Pneumothorax, lung sound mapping, polycystic lung disease**

I. Introduction

Pneumothorax has been managed by chest tube insertions as standard therapy. However if that is small one can observe without chest tube insertion specially those patients who have underlying poly-cystic lung disease. In fact one of the 2 patients received chest tube insertion but the tube was wrongly placed into one of the large bulla causing lung-pleura fistula with significant air leak requiring lengthy hospital admission with surgical procedures (which could happen if underlying bulla in the lung appears like part of pneumothorax, if you depend on chest x-rays).

Therefore we followed these 2 patients as extent of pneumothorax was 12% to 15% by X-rays using lung map of breath sounds to monitor progress of the pneumothorax.

II. Materials and methods

Two young teenage Japanese patients known to have polycystic disease of the lung who came in with spontaneous pneumothorax. Both cases were managed conservatively without attempting chest tube insertion as it could be difficult to do chest tube insertion because of presence of large bullae within the lung of these patients with polycystic lung disease.

We monitored air distribution of entire area of the lung by recording lung sounds in the entire area of the lung, producing lung map using Murphy STG-16 (microphones are placed over 1 trachea and 14 areas of lung field from apex to bottom of the lung`s lung sounds with 1 heart sound recording) while in sitting position.

III. Results

We were able to manage spontaneous pneumothorax on both patients with high risk polycystic disease of the lungs without using invasive method of management—namely chest tube insertion. Monitoring lung marks of the lung sounds by STG-16 of Murphy made it possible.

IV. Discussion

Although there was no clear sign of tension pneumothorax in the beginning there was definite fear of that besides another blebs pop open to create another pneumothorax that would be devastating to both patient`s outcome. We are lucky that was never happened needless to say monitoring of the lung sound map of entire lungs was a savor for both cases.

V. Conclusion

It became happy ending thanks to Dr. Murphy`s STG-16 (Inventor of the system and founder of ILSA)

Auscultation by conventional stethoscope is never complete to detect changes of air distribution of entire lung fields, especially some area of atelectasis caused by localized pneumothorax.

Cardiac Response to Respiration in Deep Breath-in vs. 2 Breath-in and 2 Breath-out

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Abstract— **It has been known that heart beat becomes slower on inspiration. When one takes deep breath it leads to more negative pressure within the chest to be generated. This tend to lead to more blood returned to the left ventricle, hence a delay of the next heartbeat is produced. We have been studying this phenomenon for the past 4 years and reports have been made at 36-39 ILSA on non-smokers, smokers and patients with COPD. From my observations during a cold weather, when I breathe 2 sequential inspirations, followed by 2 sequential expirations made my heart beat slower and easier to fast walk during the winter. Therefore we studied 20 never-smokers subjects to see whether these 2 sequential inspirations followed by 2 sequential expirations make the heart beat become slower in an ambient room before undertaking cold environment study. This didn`t produce any advantage over deep breathing in ambient warm room temperature environment.**

Keywords— **heart rate, inspiration, expiration, never-smoker patients**

I. Introduction

It has been known that the heart beat becomes slower on inspiration which we have been studying during the past 4 years by simultaneously recording breath sounds and electrocardiograms (QRS complex) in various subjects: Smokers, Non-Smokers and disease conditions such as COPD and Asthma. This study is involving only never-smokers of various ages, whether breathing patterns influence this delayed heart beat phenomenon.

II. Materials and methods

20 randomly selected never-smoker patients during my office visits of various conditions. Ages ranging from 20 to 69, half of them are female patients.

We simultaneously recorded breath sounds and electrocardiograms (QRS) for all 20 subjects who were on sitting position in my office.

Slow deep breathing of 10 breath and 2 sequential breath in , followed by 2 sequential breath out of 10 breath have been made each subject and 2 QRS intervals were measured in msec.

III. Results

On slow deep breathing the heart rate between inspiration and expiration was 1.17 while with 2 breaths out the heart rate change between inspiration and expiration was 1.12. For the average of all 20 never-smoking subjects there were no significant differences between male and female or various ages.

IV. Discussion

We let each subject freely to take slow deep breath and we observed that more mouth breathing is done deeper breath in most patients and get larger volume of total air moved in and out of the chest.

As I encouraged letting them breathe through both nostrils while doing 2 breaths in and 2 breaths out we noticed that the total air moved in and out of the lung was smaller in volume, partially because of the obstructed nasal canal due to the allergy season in Boston.

Although we failed to show advantage of 2 consecutive inspirations through the nose over simple deep inspiration, breathing 2 breaths in through nose should be advantageous in colder environment, as breathing air is warmer and moisturized during passage of air through nose as air volume should be larger on inspiration.

V. Conclusion

We thought that 2 breath in and 2 breath out may be advantageous to slower the heart than deep breathing but unable to show that advantage while breathing warm ambient environment.

New Approach to Evaluation of Sleep Breathing Disorders Based on the Analysis of the Sounds of Snoring and Pauses of Breathing Cycle

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Abstract — **The subject of the research was the development of software for automatic processing of signaling information aimed to obtaining meaningful sleep parameters. A key feature of the study was the need to work with a large amount of signal information recorded by acoustic receiver throughout the patient's sleep.**

Keywords — **obstructive sleep apnea, snoring, screening, breathing cycle, out-of-center testing**

I. Introduction

The vast majority of cases of sleep breathing disorders especially obstructive sleep apnea (OSA) remains unrecognized due to limitation of availability of PSG studies, cost of the diagnostic equipment, low level of the knowledge of sleep breathing disorders among physicians.

There is a need for development of simple, cost effective, out-of-center device for screening of snoring and obstructive sleep apnea.

Intensive development of devices for the recording the acoustic information and the current state of development of computer technology, characterized by high performance, large memory allows to put on the agenda of the automation of the signal processing acoustic information aimed at the release of significant sleep parameters.

II. Results

To achieve the goal it had been resolved the following tasks:

1. The analysis of approaches for obtaining and pre-processing the signal information obtained by specialized means of registration;
2. The analysis of the approaches used in the processing of acoustic information aimed at extracting relevant sleep parameters;
3. The software was developed for automatic extraction of significant sleep parameters;
4. The principles of signal information processing allow to extract quantitative and qualitative sleep parameters;
5. It was developed an interface of control and signaling information processing;
6. Due to the analysis of signaling information, problems affecting the quality of extracting significant sleep parameters were revealed.

The software package (“neuron”) allows working with the signals recorded in wav format with an arbitrary sampling frequency. Random sampling rate regime means that all constants, implemented in the program are translated according to the current value of the sampling frequency. This approach provides a certain flexibility.

One of the main problems of allocating significant sleep parameter is the choice of pretreatment methods of signal information and given the significant volume of the data signal, the method chosen should not require significant computing resources.

Very often, preprocessing signaling information directed to transfer the signal from the amplitude-time domain to the time–frequency representation. This kind of procedure is more informative, although it requires significant computational resources.

However, the temporary spectral bands can contain quite a lot of information not relevant to the task.

It is obvious that one of the significant characteristics of sleep breathing disorders is apnea, characterized by a duration and frequency of occurrence. With this in mind, we decided to carry out the analysis in the time domain. What we think should save computing resources and time to process large volume of signaling information.

On the figures showed below are the examples of activity of the “neuron” and the results of its activity – automatic analysis of the snoring and obstructive sleep apneas.

Fig. 1 demonstrates the “neuron” activity – processing of the signal of simple snoring stored in a wav format. Fig. 2 demonstrates the original signal (yellow) and the results of binarization (red) of the pattern of “neuron” activity shown on the Fig.1. The pauses of breathing cycles marked by red color are periods of expirations without flow limitation. Fig. 3 demonstrates the “neuron” activity – processing the signal of snoring and pauses corresponding to sleep apneas stored in a wav format. Fig. 4 demonstrates the results of binarization of the pattern of “neuron” activity shown on the Fig.3. Besides the assignment of expiration periods there are repetitive sleep apneas marked by red color.

Fig.2 and Fig. 4 also demonstrate different types of sleep breathing events. Different colors reflect different types of sleep breathing events classified by the process of binarization of the original signal.

Classification of the pauses of the breathing cycles and the periods of sleep apneas will allow to evaluate the sleep breathing disturbances more precisely.

III. Conclusion

We developed the new mathematical algorithm (“neuron”) for automatic analysis of snoring and obstructive sleep apnea based on the analysis of the sounds of snoring and pauses of breathing cycle.

The new method will allow to screen sleep breathing disorders in large populations. We also suggest that new algorithm should favor the development of the new classification of sleep breathing events.

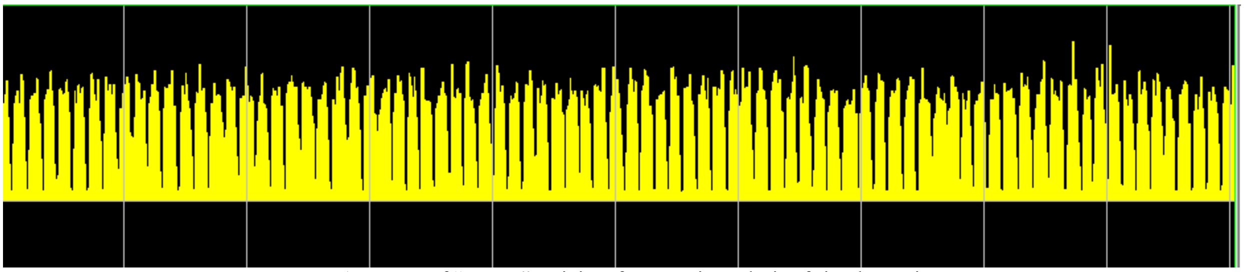
Fig. 1: Pattern of “neuron” activity of automatic analysis of simple snoring

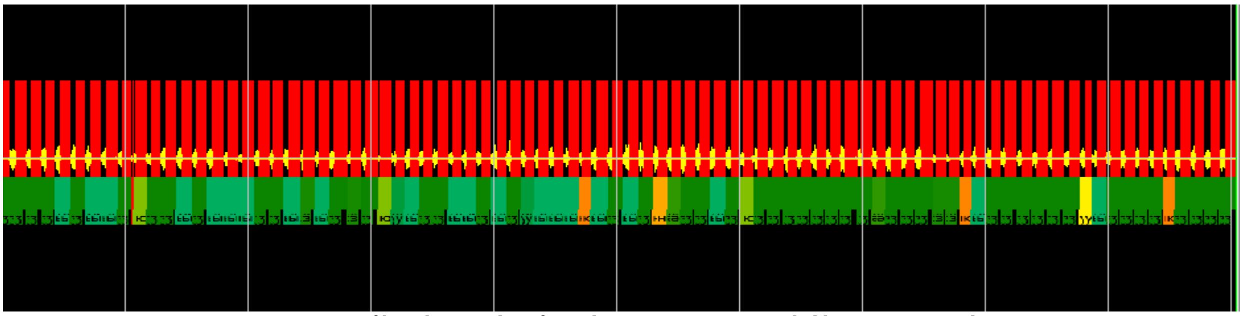
Fig. 2: Pauses of breathing cycles of simple snoring pattern revealed by automatic analysis

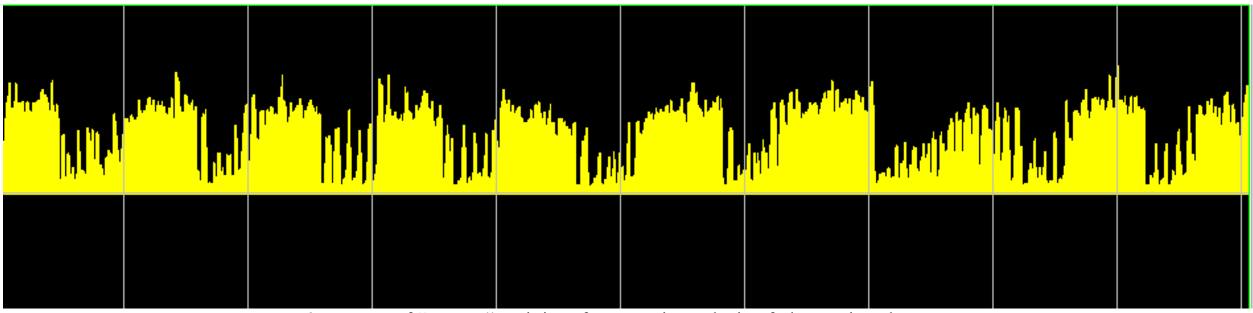
Fig. 3: Pattern of “neuron” activity of automatic analysis of obstructive sleep apneas

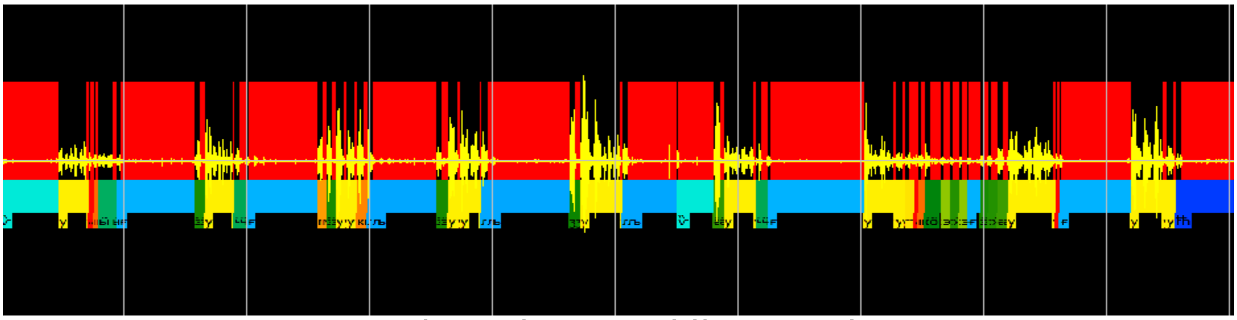
Fig. 4: Obstructive sleep apneas revealed by automatic analysis

On a Possible Mechanism of Sound Generation and Amplification in a Corrugated Tube

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Abstract— **The sound generation by an air flow in a corrugated pipe is considered in the view of instability waves theory. A simplified model of aeroacoustic interaction without feedback is used to get qualitative interpretation of this phenomenon.**

Keywords— **Corrugated tube, instability wave**

I. Introduction

It is known [1-3], that air flow in a corrugated tube (which is the shape of the trachea) can generate tonal or multitonal sound. Generation begins at a flow rate greater than a certain threshold value. The spectrum has the form of closely spaced set of peaks distributed in a narrow frequency range centered at $f_1 \approx 0.5V/l$ (V – flow rate, l – corrugation pitch). This frequency domain of generation is repeated at a higher frequency with medium frequencies of about $f_n \approx n \cdot f_1$ (n – integer).

II. Model

The theoretical explanation of the excitation tube by air flow must be based on the self-oscillating mode. A sound wave becomes instable due to some mechanism of energy transfer from flow to sound wave caused by the interaction of sound and flow with wall irregularities. Amplification of the sound vibrations occurs after aerodynamic energy pumping exceed parietal friction, heat conduction and radiation losses from the open end of the tube.

There are three working hypotheses describing transfer mechanisms: 1) a change in the impedance of the corrugation pitch under the influence of a tangential discontinuity overlying cavity; 2) generation of an axial force in inleakage oscillating tangential discontinuity, down from a previous projection of corrugation to the next ledge; 3) interaction of the eigenmode waveguide with a critical layer of flux (Miles mechanism [4]).

In all cases, the energy transfer is possible only by the processes occurring near the wall: that's where the flow has features associated with the presence of a velocity gradient (boundary layer) and the heterogeneity of the corrugated wall. This means that the generation of noise is expressed by the superficial nature and, therefore, can be described using a special form of boundary conditions at the duct walls.

The first mechanism was studied in terms of instability waves with simplified non-feedback approach. Under these assumptions the impedance Z of a single pitch is the complex oscillating function of the Strouhal number:

$$Z = \frac{\rho_0 c}{B} 2\sqrt{2}\left(2\pi \mathrm{Sh}\right)\exp\left[-2\pi \mathrm{Sh} + i\left(-2\pi \mathrm{Sh} + \frac{3\pi}{4}\right)\right],$$

where ρ_0 – air density, c – speed of sound, $\mathrm{Sh} = fl/V$. Parameter B is defined by features of running through the tangential discontinuity hydrodynamic instability wave.

The real part of the impedance can be positive or negative depending on the length of the cavity. Thus, there is a range of frequencies for which the real part of the impedance is less than zero. This area of the negative impedance is repeated at higher frequencies. If these zones of negative impedance overlap the spectrum of the natural frequencies of the waveguide length L – there is the excitement from the open end (Fig. 1). By increasing the flow rate zone of instability gradually shifts to higher frequencies, and the generated sound jumps to the next the natural frequency of the waveguide.

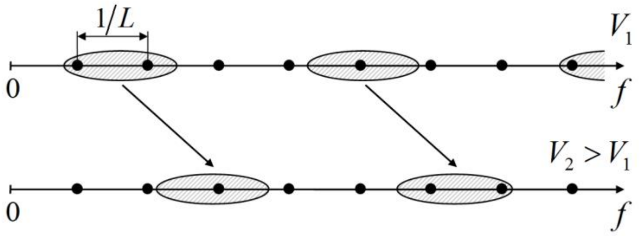

Fig. 1: Displacement of the negative impedance zones (shaded ovals) with increasing speed of flow. The black dots denote natural frequencies of the waveguide

It was shown, that this approximate approach explains main features of the sound generation phenomenon and is adjusted for $\mathrm{Sh} > 0.2$ with the exact feedback model [5].

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Acoustic Biomechanical Relationships of Forced Exhalation Revealed by Analysis of Variance among Groups with Various Degree of Bronchial Obstruction

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Abstract— **A statistically significant bidirectional impact of factor of incidence and degree of bronchial obstruction on the acoustic parameters of the forced exhalation and spirometry / bodyplethismography indicators of lung function is revealed in a sample of 218 subjects by means of Jonckheere-Terpstra non-parametric analysis of variance. It is shown that acoustic times and energies of tracheal forced expiratory noise coordinate directly with the tidal resistance and the residual volume.**

Keywords— **time, energy, noise, forced exhalation, evaluation, biomechanics, relationship**

I. Introduction

The relationships between the tracheal forced expiratory (FE) noise time FETa and spirometry indices were investigated earlier [1]. It was shown that FETa in 200-2000 Hz frequency band reflected FE bronchial resistance to a certain extent. The objective of the work is a biomechanical interpretation of the FE acoustic parameters at trachea by comparison with body plethismography and spirometry.

II. Materials and methods

Tracheal FE noises were recorded under acoustic study according to previously developed procedure [2]. Sound files were processed by previously developed algorithm to evaluate for the time FET_a in the frequency band 200 - 2000 Hz, time, normalized for age in healthy $FET_a' = FET_a / (0.78 + 0.033*Ag)$, where Ag is age, bandpass 200-Hz times $t_{200-400}$, …, $t_{1800-2000}$ and bandpass 200-Hz energies $A_{200-400}$, …, $A_{1800-2000}$.

FEV_1, FVC, FEV_1/FVC registered performing the spirometry. Bronchial resistance of tidal exhalation R_{ex} and bronchial resistance of entire respiratory cycle R_{tot}, functional residual capacity FRC, reserve volume RV, total lung capacity TLC and ratio RV/TLC were obtained by means of bodyplethismography. Actual values of the parameters were compared with predicted values to obtain the percentage % of predictions.

A total of 218 male and female volunteers, aged 16 to 68 years, were studied. The sample was splited into 5 groups – 1 – healthy, 50 subjects, age Me;LQ;UQ - (25.0;21.0;30,0); 2 – persons with risk factors of development bronchial asthma (BA) and chronic obstructive pulmonary disease (COPD), 60 subjects, age (23.0;21.0;29,5); 3 – spirometry negative asthma patients, 32 subjects, age (31.5;20.5;48.0); 4 – spirometry positive asthma patients, 41 subects, age (37.0;26.0;64.0), 5 – COPD patients, 35 subjects, age (58.0;64.0;62,0).

The incidences and degree of bronchial obstruction evidently increase from group 1 to group 5. Thus "incidence and degree of bronchial obstruction" may be treated as a factor with 5 different levels presented by the studied groups. This is a problem commonly analized by classic ANOVA. However a nongaussianity of distribution and a nonhomogeinity of analized parameters variances prevented using this method in the case. Consequently the nonparametric analysis of variance by means of Jonckheere-Terpstra test (SPSS, Polar Engineering and Consulting) which did not require the gaussianity and the homogeinity of variances was used as the basic method.

III. Results and discussion

The ordered dependence of FET_a on the factor levels is illustrated at Fig. 1. Results of calculation of values of standard normal distribution z, approximating Jonckheere-Terpstra statistics, and corresponding levels of 2-side significance are shown in the Tabs 1 - 3. It is evident that a degree of significance of the variants dependence on the studied factor is indicated by z value. However, the required relationship may be partly caused by the presence of statically significant correlation of the age of subjects with the studied factor. To eliminate this extra influence seems justified to introduce a relative indicator z/R, which is the value of the standard normal distribution, normalized to the Spearman correlation coefficient. Let's take double value of this ratio for the variant "Age" as a threshold. When there is no significant correlation under calculation of this ratio let's assume R = 0.1. The values of the ratio z/R, exceeding the threshold are highlighted in the tables with "bold".

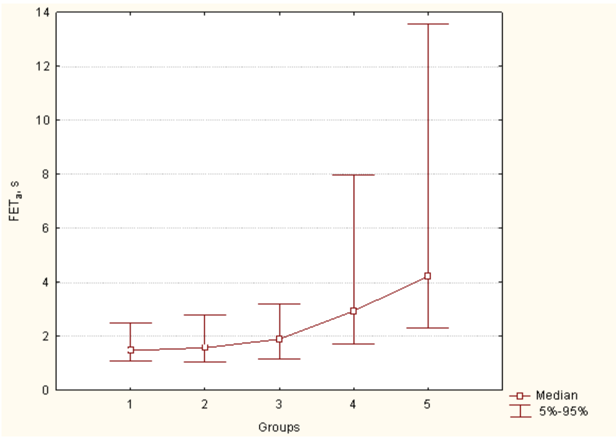

Fig. 1: Box and whiskers diagram of FET_a

Table 1: Results of analysis: age, spirometry and bodyplethysmography indexes

Параметр	Age	FVC	FEV_1	FEV_1%	FEV_1/FVC	R_{ex}	R_{tot}%	RV%	RV/TLC	RV/TLC%
z	7.64	-3.99	-8.48	-8.83	-11.68	9.62	9.44	5.72	8.09	4.36
p	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000
z/R	7.64	10.25	13.91	**35.32**	**17.96**	**25.31**	**28.6**	**35.75**	13.49	**43.63**

Table 2: Results of analysis: acoustic time parameters

Параметр	FET_a	FET_a'	$t_{200-400}$	$t_{400-600}$	$t_{600-800}$	$t_{800-1000}$	$t_{1000-1200}$	$t_{1200-1400}$	$t_{1400-1600}$	$t_{1600-1800}$	$t_{1800-2000}$
z	10.76	9.26	9.69	10.77	10.31	8.67	8.18	8.01	5.20	3.57	3.07
p	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000
z/R	**15.37**	**29.87**	**18.99**	**16.57**	**15.39**	**16.68**	**15.43**	14.83	11.56	9.92	10.24

Table 3: Results of analysis: acoustic energy parameters

Параметр	$A_{200-400}$	$A_{400-600}$	$A_{600-800}$	$A_{800-1000}$	$A_{1000-1200}$	$A_{1200-1400}$	$A_{1400-1600}$	$A_{1600-1800}$	$A_{1800-2000}$
z	4.15	4.59	6.03	5.01	4.68	5.85	3.08	1.56	2.65
p	0.000	0.000	0.000	0.000	0.000	0.000	0.002	0.12	0.008
z/R	**41.5**	**22.97**	**23.21**	**31.33**	**22.3**	**17.73**	11.4	-	12.6

It was expected, that "the incidence and degree of bronchial obstruction" increased from group 1 to group 5. Revealed significant trend of decreasing index FEV_1/FVC, which is the main indicator of the presence of airflow obstruction, in the order of numbering groups confirms the thesis (Fig. 2). Another index characterizing the severity of bronchial obstruction FEV_1% has a similar behavior (Tab. 1).

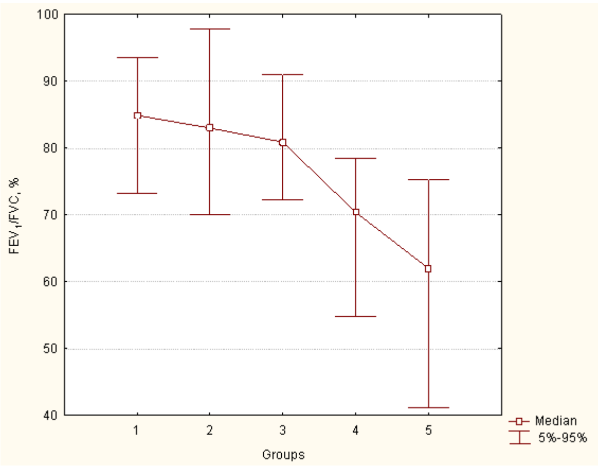

Fig. 2: Box and whiskers diagram of FEV_1/FVC

On the contrary bodyplethysmografy indexes R_{ex}, R_{tot}%, RV%, RV/TLC% are characterized by the significant trend of increase in the order of numbering groups (Tab. 1). It is not surprising since R_{ex}, R_{tot} parameters describe an airway resistance, and RV, RV/TLC parameters, named as "airtrapping" indexes, characterize a pulmonary hyperinflation. These parameters often increase under bronchial obstruction and positively correlate with its severity.

The absolute value of tracheal forced expiratory noise time FET_a, as well as normalized one FET_a', also have a significant trend of increasing in the order of numbering groups (Tab. 1). Other acoustic parameters, including a large part of 200-Hz bandpass times ($t_{200-400}$ … $t_{1000-1200}$) and energies ($A_{200-400}$ …. $A_{1200-1400}$) demonstrate similar behavior. However in more high frequency 200-Hz bandpass times ($t_{1200-1600}$ … $t_{1800-2000}$) and energies ($A_{1400-1600}$ … $A_{1800-2000}$) the significant ordered dependences disappear (Tab. 1).

Thus, with the statistical model, characterized by a significant gradual increase of airway resistance and residual lung volume, we revealed the significant growth of most acoustic parameters of tracheal FE noise. This implies that an increase in said acoustic indexes of FE represents a certain measure of the airflow obstruction as well as of the lung hyperinflation. The deduction is consistent with the previously developed model acoustic representations that the prolongation of tracheal FE noise may be caused by an increase in the total resistance of the bronchial tree, mostly in the large bronchi [2], which corresponds to an increase in R_{ex}, R_{tot}%, and delayed emptying of lung units ventilated by smaller bronchi [2], which reflects the growth of RV%, RV/TLC%.

IV. Conclusions

A statistically significant bidirectional influence of the factor of incidence and degree of bronchial obstruction on the FE acoustic parameters and spirometry/ bodyplethysmography indexes of lung function is revealed by means of nonparametric analysis of variance.

It is demonstrated that tracheal FE noise acoustic times and energies coordinate directly with both the tidal resistance, and the residual volume, measured by bodyplethysmography, that confirms the previously developed model representations about noise production under forced exhalation in healthy and patients with bronchial obstruction.

The frequency selectivity of the dependence of acoustic time and energy on the factor of incidence and degree of bronchial obstruction is found.

The study was supported by RFBR grant 14-04-00048.

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On Estimating a Wheezing Source Distance by Means of Intensimetry Processing of Sound Responses Recorded above the Chest

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Abstract— **The set of equations for determining the distance from the chest surface to the wheezing sources of different types (monopole, dipole, transverse quadrupole) in human lung is obtained. Under trial implementation anatomically plausible estimates of distances to wheezing sources of frequency range 100 - 550 Hz are experimentally determined. It is demonstrated the possibility of resolution of distances to sources having different peak frequencies.**

Keywords— **wheezing source, range, intensymetry, monopole, dipole, quadruple**

I. Introduction

A localization of sources of wheezes in human lungs is an actual medical diagnosis problem. Using the intensimetry approach for processing respiratory sounds recorded on the chest surface makes it possible to estimate a distance from chest surface to wheezing sources in human lungs [1].

The objective is detailed study of this method.

II. Materials and methods

We developed combined acoustic sensor (CAS), shown at Fig. 1. The simulation and numerical estimates showed that for frequencies above 100 Hz the response of CAS microphone channel at chest surface is proportional to oscillatory displacement in longitudinal sound wave [2]. Whereas the response of the ring piezotransducer is proportional to dynamic force (contact pressure).

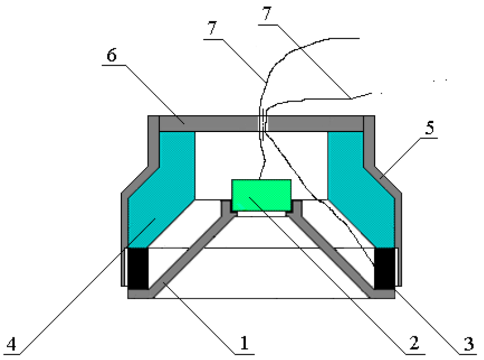

Fig. 1: Combined acoustic sensor
1 – stethoscope head, 2 – microphone, 3 – ring longitudinal piezotransducer, 4 – massive ring pad, 5 – housing, 6 – cover, 7 – cables

Most of known models of wheezes formation predict their generation in limited by the length part of the airway (AW) which is of an order of 1 cm. The diameters of the AWs, in which wheezes are most likely produced lie in a range of 2-10 mm. Since the speed of sound in the lung parenchyma is about 30 m/s the length of longitudinal sound waves with a frequency of 300 Hz is near 10 cm. Thus a wheezing source can be considered as dotted in the wave sense. A dotted source of sound can vibrate in different types - as monopole, dipole or quadrupole.

Let's consider a dipole wheezing source [3] with random amplitude and fixed orientation, which emits sound in the lung parenchyma tissue structure – a medium with low shear modulus and high damping. Its sound pressure and radial component of oscillatory velocity may be presented as

$$p = B * exp(- ikr)/r * [1 - i/kr] * cos\varphi,$$
$$v_r = B * exp(- ikr)/\rho cr * [1 - 2i/kr - 2/(kr)^2] * cos\varphi,$$

where B – dipole constant, k – wavenumber, r – distance between the source and CAS, ρ – medium density, c – sound speed, φ - azimuthal orientation of the dipole.

Returning to the representation of the CAS channels as a dynamic force receiver and the receiver of the oscillatory displacement with accuracy up to the sensitivity of the receivers [1], and calculating the ratio of the real and imaginary parts of the cross spectrum W, we obtain $Re(W)/Im(W) = [1/kr + 2/(kr)^3]$.

Assuming experimentally measured value of ratio of real and imagine parts of cross spectrum for peak frequency f $Re(W)/Im(W) = C$, and introducing designation $z = 1/kr$, we obtain the equation

$$2z^3 + z - C = 0 \qquad (1)$$

Similarly, in [1] the equations were obtained for the source in the form of a monopole and a transversal quadrupole

$$45z^5 + 12z^3 + 3z - C = 0, \qquad (2)$$
$$z - C = 0. \qquad (3)$$

If the average speed of sound in the common medium – lung parenchyma – chest wall is c, then solving equations (1-3), we obtain a distance $r = c/2\pi fz$ for wheezing sources of different types.

III. Results and discussion

Registration of breath sounds of inspiration and forced exhalation was performed in healthy volunteer. CAS was placed in the right subscapular region. Recording was made with sampling frequency 10 kHz and 16 bit dynamic range with electronic recorder PowerLab (ADInstruments). Spectrograms of CAS channels were calculated. The paths in spectrogram corresponding to wheezes were identified.

Cross spectra of CAS channels responses were calculated for time domains containing the identified paths. Wheezes with peak frequencies 341.8, 498, 537.1, 1152.3 Hz were found in the most powerful part of forced expiratory noises. More weak wheezes with peak frequencies 1328.1 и 1757.8 Hz are revealed in the end of forced exhalation. During inspiration weak wheezes with peak frequencies 175.8, 234.4, 322.3 were identified. Values of *Re(W)*, *Im(W)* for these peak frequencies were measured in linear scale and parameters *C* were calculated. Positive real roots of equations (1-3) were numerically evaluated and the corresponding distances/wavelengths were estimated (Tab.1).

Table 1: Distance (*r*) [cm] / wavelength (*kr*) from CAS to the wheezing source

Frequency, Hz	*r/kr*			*Re(W)/ Im(W)*
	Quadrupole	Dipole	Monopole	
Powerful forced expiratory wheezes				
341.8	**16.9/12.1**	6.1/4.3	5.5/3.92	0.255
498	**16.4/17.1**	5.7/5.9	5.4/5.62	0.178
537.1	**16.4/18.5**	5.7/6.4	5.4/6.09	0.164
1152.3	*28.3/62.2*	*9.5/22.8*	*-9.4/2.27*	*-0.044*
Weak wheezes in the end of forced exhalation				
1328.1	*2.1/5.8*	*0.84/2.3*	*0.62/1.72*	0.582
1757.8	*1.4/5.2*	*0.58/2.1*	*0.4/1.47*	0.681
Weak wheezes during inspiration				
175.8	22.6/8.3	8.5/3.1	**7.1/2.61**	0.383
234.4	15.5/7.6	6.0/2.9	**4.8/2.37**	0.422
322.3	14.3/9.6	5.3/3.6	**4.6/3.07**	0.325

Notes: the solutions implausible for physical or anatomical reasons are marked as italics, the most plausible solutions are marked in bold.

Since a chest wall thickness is about 2-3 cm the solutions, giving lower distances may be discarded from anatomical considerations. Negative value of *Re(W)/Im(W)* for wheeze with peak frequency 1152.3 Hz is in obvious contradiction with our model.

In regard of remaining solutions (Tab. 1) one can see the following features. The most powerful mid-frequency (in the band of about 400-600 Hz) forced expiratory wheezes in healthy individuals are produced primarily in the central parts of the bronchial tree (the lower part of the trachea and main / lobar bronchi). From this perspective, the distances estimated for the sources of powerful forced expiratory wheezes at frequencies 341.8, 498, 537.1 Hz when using the model of a quadrupole source, look quite plausible. Indeed, the direct distance measured with pelvis meter between the CAS position under the angle of the right scapula and the jugular depression is about 23-24 cm. Consequently, estimated values of the direct range 16.4-16.9 cm indicate location of their, apparently common, source at 6.1-7.6 cm in deep (and down) of the chest from the jugular. In accordance with anatomical reasons this location corresponds well to the bifurcations of the trachea and main/lobar bronchi. Whereas dipole and monopole source models provide almost three times lower distances from the source, thus considering unrealistic in this case.

In contrast, inspiratory breathing noises in healthy individuals presumably are formed in much more distally disposed parts of the bronchial tree – into 9th-13th levels of bronchial tree [1]. Therefore, distances 14.3-22.6 cm, obtained here from the equation (2) for the quadrupole emission seem to be fabulously large. This conclusion is reinforced by the fact that the noises of inspiration are of much less power than forced expiratory noises, and it is unlikely that they could be transmitted by lung structure from so remote from the sensor part of the chest due to high sound attenuation in the lung parenchyma. Distances obtained for the monopole and dipole sources look much more realistic in this case. According to [1] for the inhalation noise monopole character of emission is the most anticipated. For this type of source the calculated distances allow to allocate two sources (Tab. 1) – one at the distance of 7.1 cm, the second – 4.6-4.8 cm. However the dipole source model gives in this case sufficiently close estimates. It is interesting to note that here two sources with different peak frequency of wheezes can be resolved for all considered models of emission.

Thus for 6 of the 9 registered wheezes using the proposed method can obtain plausible estimates of their distances from the source to the surface of the chest. However there are basic limitations of the developed method, which involve the feasibility of assumption on structural transmission of the sound in lung parenchyma, the feasibility of assumption on dotted character of the source, the heterogeneity of the velocity of sound in the lung tissue, an ambiguity of the choice of models of the emission source. Consequently further investigations are welcome to detail possibilities and limitations of the approach.

IV. Conclusions

The set of equations for determining the distance from the chest surface to the wheezing sources of different types (monopole, dipole, transverse quadrupole) in human lung is obtained. Under trial implementation anatomically plausible estimates of distances to wheezing sources of frequency range 100 - 550 Hz are experimentally determined. It is demonstrated the possibility of resolution of distances to sources having different peak frequencies.

The study was supported by the RFBR grant 13-08-00010 and the 2015-2017 scholarship grant of the President of the Russian Federation for young scientists and postgraduate students.

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A Comparative Analysis of Acoustic Sensors for Recording Respiratory Sounds at the Chest Surface

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Abstract— **The types of acoustic sensors for recording respiratory sounds at the chest surface are described. Mechanisms of their functioning are shown. Features of the sensors are compared.**

Keywords— **acoustic sensor, chest surface, types, mechanisms, noise, parameters.**

I. Introduction

Acoustic sensors for recording lung sounds at the chest surface should meet some basic requirements. These requirements apply to achievement the acceptable sensitivity (sensitivity threshold) to the useful signal, sufficient noise immunity against noises, and a linearity of frequency response. Specific conditions of recording acoustic waves propagating through human thorax consist in its registering on the border of the body with air medium. This border can be regarded as an acoustically soft. Existing types of acoustic sensors used here may be divided into contact and non-contact receivers.

II. Results and discussion

Non-contact sensors are to some extent exotic, so let's start with the consideration of contact sensors.

Currently, three types of acoustic sensors mounted in contact with the chest surface are used to record respiratory sounds – acoustic accelerometers (Fig. 1), stethoscopic sensors with microphones (Fig. 2), and so called "contact" sensors (Fig. 3), in which the sensitive piezoelement is situated between the surface of the chest and the housing.

Any acoustic sensor having a mass M when placed on a layer of soft tissues (skin and adipose layer) having hardness K should inevitably have some eigenfrequency of suspending, which in the long wave approximation of a rubber vibration damper, as well as with a small viscosity of biological tissues may be found as $f_0 \approx (K/M)^{0.5}/2\pi$. With M $\approx$ 7-8 g, and sensor diameter of about 30 mm the f_0 is assessed for the lower chest surface being about 200 Hz [1].

When operating at frequencies 1.5–2 times lower than f_0, the sensor will make common vibrations with the surface of the chest. On the contrary, when operating at frequencies substantially higher than f_0, the sensor can be considered isolated from vibrations of the medium. Note that the linear vibration isolation effect, according to the known law $20\log(f/f_0)$, begins to manifest itself from frequencies of about $(1.5–2)f_0$ and increases with elevating frequency.

Let's consider the accelerometer (Fig. 1). Such an acoustic sensor if the resonance of its sensitive element is higher than the analyzed range, during operation at frequencies lower than $f_0/(1.5–2)$, will make common vibrations with the surface of the chest and, thus, will be a classic vibrational acceleration receiver. During operation above f_0 (taking into account the low Q factor of suspending resonance), such sensor will be a vibrational velocity receiver, owing to compensation of the accelerometer's natural frequency response, by an almost linear drop in the transfer characteristic due to vibration isolation effect, beginning from approximately $(1.5–2)f_0$ [1].

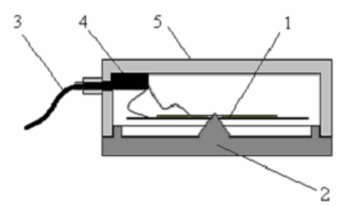

Fig. 1: Accelerometer sensor: 1 - round bimorphic piezoplate, 2 – base, 3 - cable, 4 – preamplifier, 5 - housing

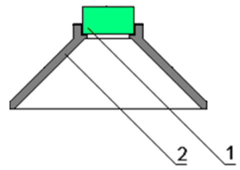

Fig. 2: Stethoscopic sensor: 1 - microphone with preamplifier, 2 - stethoscopic head

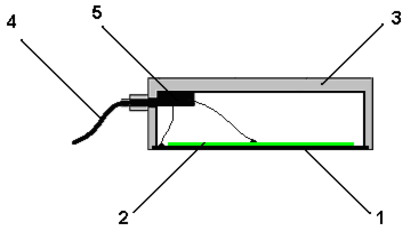

Fig. 3: "Contact" sensor with bent piezoelement: 1 - elastic plate, 2 - round piezoplate, 3 – housing, 4 - cable, 5 - preamplifier

The stethoscopic sensor with electret microphone built into the bell (Fig. 2) above $(1.5–2)f_0$ works as vibrational displacement receiver, while below $f_0/(1.5–2)$ it has very low sensitivity due to high vibrational immunity of the microphone [1].

The "contact" acoustic sensor (Fig. 3) below $f_0/(1.5–2)$ works as accelerometer while above $(1.5–2)f_0$ it has a characteristic of dynamic force (contact pressure) receiver, which in locally plane wave in the first approximation can also be considered as vibrational velocity receiver [1].

Idealized frequency responses of sensors with assumption of equal masses and diameters are represented in Fig. 4. One can see that the real value of the suspending eigenfrequency f_0 may be decreased for all three types of sensors by elevation of mass to elongate the working linear range in $f>f_0$ band. While the working linear range in $f<f_0$ band may be extended partly only for the accelerometer sensor by means of its mass decrease. However this expansion is strictly limited at 200-300 Hz by square root

low $f_0 \approx (K/M)^{0.5}/2\pi$ and achievable mass of modern accelerometers having acceptable sensitivity threshold.

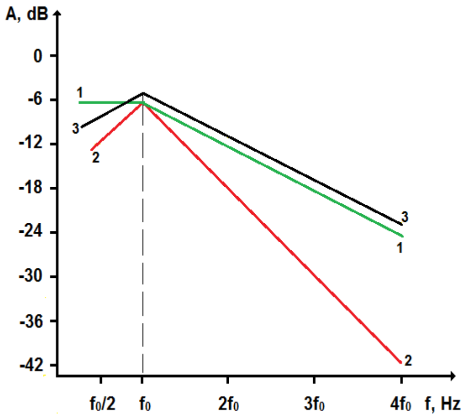

Fig. 4: Idealized curves of frequency response of sensors at the chest surface – 1 – accelerometer, 2 – stethoscopic receiver with the microphone, 3 – “contact” receiver

Thus the best characteristics in terms of mechanic-acoustic conversion efficiency and the absence of frequency response distortions in the range of 100–700 Hz for recording respiratory sounds on the chest surface belong to the heavy stethoscopic receiver with a microphone and the heavy “contact” receiver with a piezotransducer, the resonance frequency of which lies substantially higher than the studied frequency range. There is a difference in the frequency slope of these sensors, making “contact” sensors more pertinent for recording high frequency sounds than stethoscopic sensor.

The light accelerometers as well as the light “contact” sensors with flexural sensitive piezoelement have susbstantial distortions of the frequency response in the range of 100–700 Hz. However the light accelerometers with high resonant frequency are the only suitable for recording sound in the high frequency range of 10-19 kHz.

It should be noted that all types of sensors are exposed to noises and their noise immunity is different. The most frequent noises are external sound signals, propagated through air medium; vibrations involved by oscillations of the chest surface or operator’s hand, holding sensor; and so named pseudo sound, caused by oscillatory compression-expansion of elastic volume, adjacent to sensitive element. The last kind of noise is connected with mechanical vibrations.

The stethoscopic receiver with the microphone (Fig. 2) has low immunity to external sound signals, as well as pseudo sound, while its immunity to vibrations is high. The accelerometer (Fig. 1) is highly sensitive to vibrations, whereas its immunity to external sound signals and pseudo sound seems remarkable. The “contact” receiver has acceptable noise immunity to external sound signals; however it is prone to vibrations/pseudo sound.

The combined acoustic receiver, containing a heavy “contact” sensor and a stethoscopic sensor with electret microphone, which responses are intensimetry processed, seems a promising solution in terms of noise immunity [2].

Another problem is multichannel recording. When sensors are placed “back to back” at chest surface a cross influence on their responses could be essential through damping chest wall oscillations and exiting surface waves. The best performance in the case is achievable by light accelerometers, while heavy “contact” sensors are the worst.

As for non-contact sensors, only optical receivers have acceptable sensitivity threshold now.

First of them are the laser interferometer oscillatory displacement receivers or the Doppler oscillatory velocity receivers. There are the scanning versions now making possible serial pick up of signals in various part of chest [3]. Such sensors are ideal for multichannel recording (only light reflecting film should be mounted at chest) and their $f_0 \rightarrow 0$. However these sensors have some disadvantages too. The device is not portable and is highly expensive. The features of frequency response in combination with sensitivity threshold still limit their high frequency range at about 300 Hz [3]. Though the receivers are non-sensitive to pseudo sound, the problem of their immunity to vibrations of chest as well as apparatus, and external sound signals remains poorly understood.

Another type – videogrammetry optical sensors have similar disadvantages, exacerbated by a low dynamic range.

It looks like there is no one optimal sensor for all scenarios of acoustic investigation of lungs. Thence a passive recording of lung sounds at chest is more frequently performed by a stethoscopic receiver with microphone (especially at law frequencies) or a “contact” sensor. Whereas recording under transmission sounding is performed better with an accelerometer or “contact” sensor.

III. Conclusions

There is no an optimal sensor for all scenarios of acoustic investigation of lungs. All known sensors have certain advantages as well as disadvantages in their usage. New studies on designing optimal sensors are welcome.

Acknowledgments

The study was supported by the RFBR grant 13-08-00010 and the 2015-2017 scholarship grant of the President of the Russian Federation for young scientists and postgraduate students.

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Monitoring Physiologic Status of a Diver by Means of Respiratory Sounds Recorded under Diving Suite *in Situ*

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Abstract — **By means of registration of breath sounds underneath wetsuit it is possible to measure the breathing rate of divers under water without distorting the paths of breathing apparatus both open and closed circuit types. Open circuit scuba generates a wideband noise, which is 25-35 dB above the noise level of closed circuit rebreather scuba in the high frequency (1 kHz), making the determination of breathing rate more reliable. The breathing rate and its variation can be an indicator of condition not only for respiratory system, but also for other systems of the body.**

Keywords — **diver, breath sounds, voice recorder, inter-breath intervals, breathing rate variability, scuba**

I. Introduction

The objective control of diver's state under water is an important but unsolved problem, especially in recreational diving. Monitoring of diver's state can be done in regard of main body systems, such as nervous, cardiovascular and respiratory. In this paper the parameters of respiratory system are investigated.

II. Materials and methods

Diver's equipment consisted of a wetsuit and a self-contained underwater breathing apparatus (scuba). Two types of breathing apparatus were used – closed circuit rebreather (CCR) scuba and the open-circuit-demand scuba where the diver breaths compressed air and exhales into the water. The registration of breath sounds was performed with H1 recorder (ZOOM Corporation, Japan). Uncompressed format WAV was used: 24-bit quantization, sampling frequency of 44.1 kHz. The sampling frequency was reduced to 8 kHz under processing by the software. Low-pass filtering and automatic gain controling were disabled. The recording was done to the external memory (32GB microSDHC card). Built-in stereo microphone was used as a sensor. The recorder had an adjustable input amplifier 0 - 39 dB. The amount of put-up gain coefficient was chosen individually. The recorder was placed into a sealed three-layer elastic latex shell. The thickness of each layer of the shell was 0.06-0.08 mm.

The packed recorder was placed on the area of the jugular fossa and was pressed to the body by the wetsuit. The material of the suit is porous synthetic rubber (Neoprene) 5-mm thick, covered with nylon. The suit separates the shell of the recorder from the outside environment. In this condition the latex layer acts as a membrane sensitive to fluctuations of the environment pressure under the diver's wetsuit. The fluctuations are passed to the sensor that is built-in microphone through the airspace inside the shell. The organization of hearing of fish by swim bladder is similar to this process by its action [1,2].

Breath sounds were recorded during the immersion sessions. Water temperature was 20-22ºC. Submergences were carried out in the same coastal area, acoustic and weather conditions. The divers performed various maneuvers for one hour.

III. Results and discussion

The recorded signals were processed with LabChart (ADInstruments, Australia) software program. The recorded breath sounds of the diver with CCR band pass filtered at the frequency of 150-200 Hz are shown in Figure 1, Channel 2.

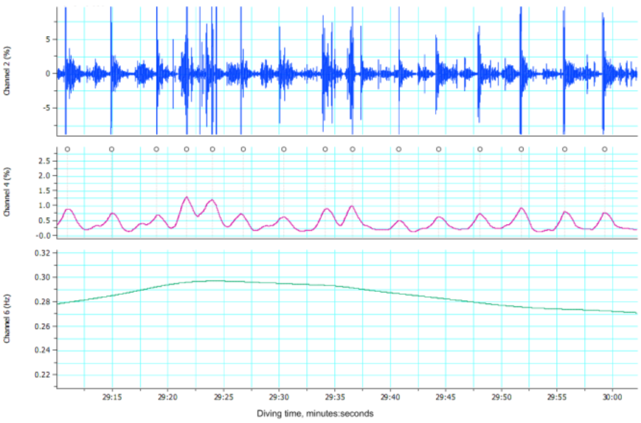

Fig. 1: Timing diagrams of CCR scuba diver: Channel 2 – oscillogram of breath sounds, Channel 4 – detected signal of breath sounds, white index mark at the top - the moment of maximum expiratory signal, Channel 6 - periodogram of inter-breath interval. Ordinate: Channel 2 and 4 - the percentage of the full scale from recorder signal, Channel 6 - frequency, Hz

The signal after linear detection and smoothing by triangular window with width 1 second is shown in Fig. 1. Sound spectra are shown in Fig. 2, 3. To calculate an average spectrum of inhalation in a 45-second recording operator drowned out all exhalations and built a spectrum (fft 1024, Hanning). Similarly, to calculate an average spectrum of exhalation all inhalations were brought to the zero level. The average range of a pause was built for a 5-second segment, when the diver holds his breath. The duration of 5 seconds does not provide such an important smoothing of the spectrum as the 45-second segment, of course.

The difference between spectrum (Fig. 2) and spectrum (Fig. 3) can be attributed to the different sound sources for breathing equipment of open and closed circuits types. For instance, open circuit scuba generates a wideband noise. Its noise spectrum at 1 kHz exceeds the level of noise of CCR scuba for 25-35 dB. At the phase of inspiration the noise is generated by a scuba regulator (1st, 2nd stage). While at the phase of exhalation it is produced by floating-up bubbles with a delay at the end of exhalation. This masks the pause between exhalation and the following inhalation. During the pause heartbeats are audible, which is due to the proximity of the recorder to the field of venous pulse. On the contrary CCR scuba noises has perceptible pauses both before inhalation and before exhalation.

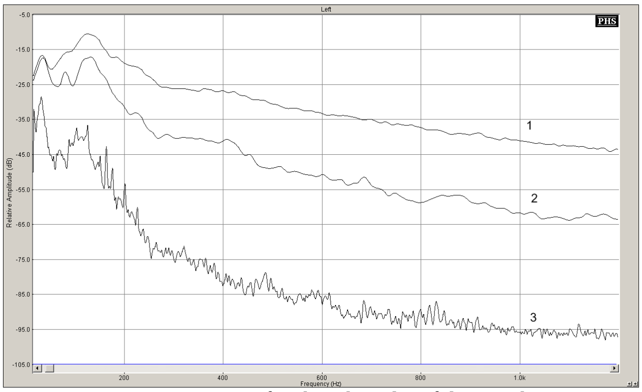

Fig. 2: An average spectrum of 12 breath cycles of diver with open-circuit-demand scuba: 1 - exhalation, 2 - inhalation, 3 – 5-second pause between inhalation and exhalation

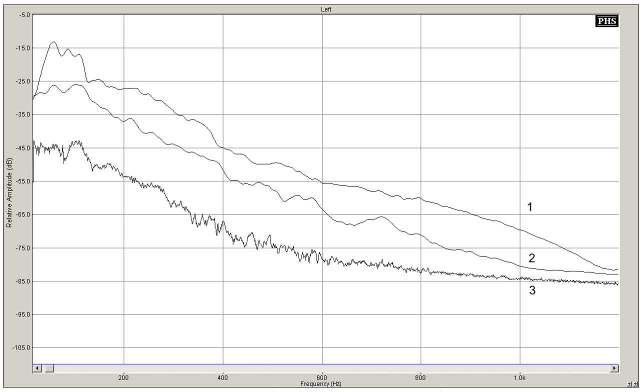

Fig. 3: An average spectrum of 12 breath cycles of diver with CCR scuba: 1 - exhalation, 2 - inhalation, 3 - pauses between inhalation and exhalation

The inter-breath intervals have been identified (Fig. 1, Channel 4, white marks on the top) by Cyclic Measurements dialog (LabChart) with detection settings – Sine shape, Minimum peak height – 1 Standard Deviation. The fragment of breath frequency dependance on time during immersion is shown in Fig. 1, Channel 6. One can see that it may be assessed with acceptible accuracy and high resolution in time.

Thus, recording breath sounds is applicable to measure submerged diver's respiration rate. It’s noteworthy that such measurements may be made without distorting the paths of breathing apparatus. The proposed method and apparatus can be used to monitor diver's condition. For these aims parameters of analyzing of breath rate are suitable. Various shape of distribution of power noise spectral density (Fig. 2, 3) makes it possible to distinguish inhalation and exhalation phases.

An analysis of heart rate variability is well described in the literature. In a number of studies [3, 4] this method is applied to the variability of breath rate. As a tool for analyzing variability of breath rate we used HRV module of LabChart software (ADInstruments, Australia). The spectrum of 45-minute recording of breath rate of a diver with open circuit scuba is shown at Fig. 4. The maximum of the spectrum is at the frequency 0.0025 Hz.

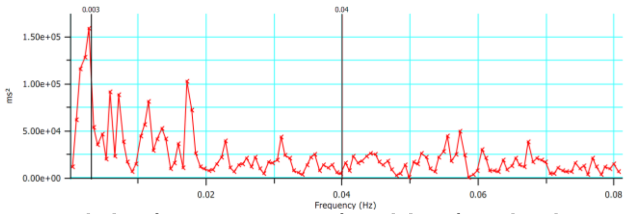

Fig. 4: The low-frequency spectrum of variability of inter-breath intervals of diver with open circuit scuba

In the paper [5] it is hypothesized that the level of the spectrum in the band of 0.002-0.0075 Hz is determined by the mechanism of metabolic control. In other words, it is related to the rate of oxygen consumption and carbon dioxide emission [3, 4]. Thus the spectral data may be applied for monitoring these metabolic mechanisms in submerged divers.

IV. Conclusions

By means of registration of breath sounds underneath wetsuit it is possible to measure the breathing rate of divers under water. This is made without distorting the paths of breathing apparatus both open and closed circuit types. Open circuit scuba generates a wideband noise, which is 25-35 dB above the noise level of closed circuit rebreather scuba in the high frequency (1 kHz), making the determination of breathing rate more reliable. The breathing rate and its variation may be an indicator of condition not only for respiratory system, but also for other systems of the diver’s body.

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Method of Pulmophonography, Created by L. Nemerovsky

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Abstract— **The 30-years history of the pulmophonography (PPG) is shown. The method and apparatuses developed by L. Nemerovski are described. The positive properties of PPG and the problems that need to be resolved are noted. The achievements of modern engineers in the development of the method and their perspectives are evaluated.**

Keywords— **pulmophonography, phonopulmograph, phonopulmogram, L.I. Nemerovsky.**

I. Introduction

Engineer L. Nemerovsky was author of about 80 different patented inventions on methods and devices for respiration support, the most significant of which is "Method for analysis of functional properties and pathological changes in the lung" (USSR Patent # 129788, co-authors L. Mishin and M. Sobakin). The method was named "pulmophonography" (PPG) and is classified as an active indirect local method of acoustic lungs study.

For objective diagnostic findings by the PPG-study of local changes in the lungs and airway conductance determination, their registration is carried out by changes in the intensity and spectra of the sound into the lungs through the upper respiratory tract and taken from different points of the chest.

II. Content

Method and device for its implementation were developed by the patented inventions of USSR:

- # 199327 "Method of the respiratory system study" (simultaneous registration of phase parameters of sound vibrations on the body surface and of the respiratory act dynamics);
- # 199328 "Device for of the respiratory system study" (implementation of the method # 199327 by spirometer using);
- # 192366 "Method of the respiratory system study" (measurement of the respiratory system complex impedance);
- # 192367 "Device for the respiratory system study" (included the digital, oscillographic and recording units);
- # 200719 "Method of lung tissue study" (excitation of trachea tissue by focused acoustic vibrations and measuring their levels in different parts of the chest).

The invention # 309699 "Method of functional properties and pathological changes in the lungs study" describes the ability of PPG-study for patients during mechanical ventilation, which is implemented by invention # 401351 "Device for the respiratory system study".

The invention № 332820 "Device for acoustic study of lungs" patented system of a visual display of the airways and lung tissue by light indicators which corresponds to the positions of sound receivers; # 332821 "Device for collection of acoustic information from segments of the chest" patented a system, comprising a lot of sound receivers with fixed positions on the surface of the chest, designed as a waistcoat or elastic chamber.

These suggestions have been implemented in the devices Phonopulmograph and Phonopulmoskop, produced by the "Medapparatura" (Kiev) since 1979 (Fig. 1).

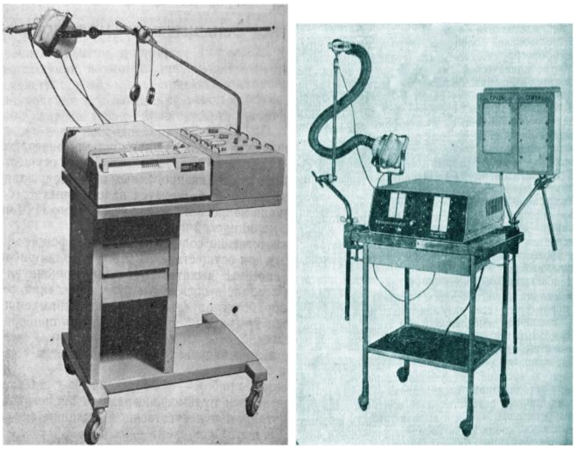

Fig. 1: Phonopulmograph and Phonopulmoskop

Named inventions have formed a PPG-technology: feeding into the lungs of the acoustic signal (frequency of 80 Hz and sound level to 60 dB). The signal that has passed through the air of lungs, partially absorbed by lung tissue and modulated by breathing, was perceived by sensors on the chest surface and was registered as a pulmophonogram (Fig. 2).

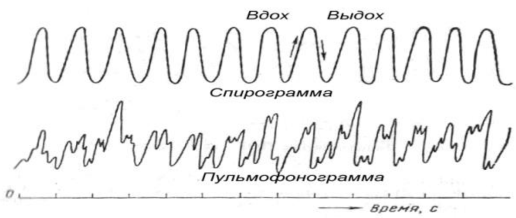

Fig. 2: Pulmophonogram of spontaneous breathing

L. Nemerovsky began to develop the theory of the sound propagation in the lungs, involving acoustic modeling techniques and structure of the bronchi (see an article "Physical and technical substantiation PPG-method study of lungs" (1966) and the invention # 249549 "Model of the tracheobronchial tree"). Using in their research this model, the author published a lot of theoretical works:

1. On the definition of the amplitude-frequency characteristics of the air tract of lungs (1968)

2. On the distribution of sound vibrations in the air tract of lungs (1968)

3. On the calculation of the parameters of the local breathing according pulmophonography data (1971)

4. Research of mechanism of pulmophonogram formation (1974)

5. On the distribution of sound vibrations on lungs (1976).

These works, complemented by experimental data, formed the basis for the monograph "Pulmophonography" (1981), comprising 7 sections:

1. Structure of the lung air tubes
2. Review of the local lungs study methods
3. The mechanism of pulmophonogram formation
4. The rules of sound distributions through the air tract of the lung
5. Methods of pulmophonogram processing
6. Trials of PPG-study in the clinic
7. Instrumentation of Pulmophonography.

Monograph defines the main advantage of the PPG-method - its adaptation to lungs physiology, the absence of use limitations in different situations; problem of method - the difficulty of interpreting the study results. To solve this problem it is need to find the relationship between PPG-parameters and local lung function. For this target, a study of the distributions of sound in airways on its models was undertaken. Local, regional, integral parameters of pulmonary function and the geometric characteristics of the lower levels bronchi, received via PPG, were used to describe the state of an air tract sections in the spontaneous and artificial breathing, controlling the dynamics of pulmonary ventilation and respiratory conditions changes.

Developed by theory and practice of PPG requirements to its equipment began principles of phonopulmographs design and the order of functional parameters calculation - the basis of algorithms for processing PPG-information.

One of the first institutions that implement the PPG-method in the clinical practice was the Moscow Research Institute of Tuberculosis. Employees of the Institute have developed some guidelines on the use of PPG, and in 1985 at the Institute there was All-Union Conference on the methodology and equipment of PPG.

L. Nemerovsky continued to work on the scientific substantiation of a method PFG. The result of this work was the the Doctor of Biological Sciences degree in 1988.

The latest invention of L. Nemerovsky in the field of acoustic lungs study was # 1836050 "Device for PPG-research" published in 1993 after his death. The purpose of the invention - the expansion of the diagnostic capabilities by recording of the amplitude-frequency characteristics of the bronchial tree branches. Interest of engineers to PPG begins to emerge from the mid-2000s. In Kharkov National University of electronics, from 2004 to 2010, Prof. H. Mustetsov and Dr. O. Vyunnik are published a lot of articles on the subject of computer processing and modeling PPG-signals and received a patent for the invention of Ukraine.

In 2010 O. Vyunnik received a PhD degree for the thesis "System of PPG-studies" (2010). The author obtained the following results:

1. Electric model of lungs with all units of the acoustic tract: the mouth, trachea, bronchi, parenchyma, pleural cavity, chest tissues to study the relationship of PPG-indices with the condition of the lung tissue and pleural cavity.

2. The method of PPG improved by analysis of all pulmophonogram components, allowing differential diagnosis of lungs pathologies.

3. Numerical simulation of changes in the intensity of sound pressure in the pulmonary segments displaying air-filling of lungs.

4. Proposed the diagnostic criteria of the lungs state based on amplitude-frequency characteristics of the signal, recorded on the surface of the chest at PPG-linking research result to the functional state of the lungs segment.

5. Hardware and software of the system for PPG-studies and processing pulmophonograms, using a computer technology.

The developed system for PPG-studies has found application in the Institute of General and Emergency Surgery and Institute of Therapy of Ukrainian Academy of Medical Sciences.

III. Conclusion

It becomes apparent a formation of a new promising area of theoretical and applied researches - respiratory acoustics, using digital imaging technology and visualization of the results of study and clinical experience in the diagnosis of the respiratory tract physiology. In this direction of research PPG can take its rightful place.

It is clear that progress in this area is not possible without understanding the nature of the sounds, not only generated by breathing, but also modulated by breathing as at PPG. Modern design can give a new impulse to the registration procedures and the interpretation of sounds. For this, the study in PPG should be focused on solving the problems of the hardware and software creation for lungs diagnostics, digital signal processing, spectral and wavelet analysis.

The Correlation between Forced Expiratory Tracheal Noise Parameters and Lung Function Characteristics in Healthy, Asthma and COPD Patients

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Abstract— **Tracheal forced expiratory acoustic parameters in the whole sample correlate more strongly with aerodynamic resistances and velocities than with volumes, except for residual volume and its ratio to total lung capacity. The correlations of opposite direction between acoustic parameters and bodyplethismography / spirometry indices in groups of asthma, COPD patients and healthy are revealed.**

Keywords— **correlation, forced exhalation, time, energy, band pass, bodyplethysmography, spirometry**

I. Introduction

Forced expiratory (FE) tracheal noises time assessment is an effective instrument of discrimination healthy subjects and young asthma patients [1]. Earlier we investigated the relationship between forced expiratory tracheal noises time (FET_a) and lung function parameters obtained by spirometry. A correlation between FET_a and lung function parameters evaluated by bodyplethysmography have not been studied yet. Moreover new forced expiratory tracheal noises parameters have been developed.

The objective is correlation analysis between extended row of FE acoustic parameters and bodyplethysmography / spirometry lung function characteristics.

II. Materials and methods

The analysis was carried out in a sample of 230 people, aged (Me;LQ;UQ) – 30;22;53. There were following groups – healthy nonsmokers, (Me;LQ;UQ) – 22;21;30 (n=50); patients with risk factors of asthma and COPD, (Me;LQ;UQ) – 23;21;29 (n=60); asthma patients with normal lung function, (Me;LQ;UQ) – 31;20;48 (n=32); asthma patients with obstructive ventilatory defect, (Me;LQ;UQ) – 37;25;55 (n=40); and COPD patients, (Me;LQ;UQ) – 58;54;62 (n=38).

Spirography and bodyplethysmography were performed with the MasterScreen Body (Erich Jaeger, Germany) apparatus in accordance with the ATS/ERS 2005 recommendations.

Acoustic examination included recording forced expiratory tracheal noises by the method described [2]. The FET_a, time in 200 Hz bands of frequencies $t_{200-400}$ ... $t_{1800-2000}$ and 200 Hz bands of energies $A_{200-400}$ ... $A_{1800-2000}$ in 200 Hz were assessed by special software.

Nonparametric Spearman rank correlation r was used.

III. Results and discussion

In the whole sample (n=230) acoustic parameters have inverse correlations of moderate to strong strength with volume velocity parameters – forced expiratory volume in 1 second (FEV_1), maximal midexpiratory flow rate ($MMEF_{25-75}$), and FEV_1 ratio to forced vital capacity FVC (FEV_1/FVC). Moderate correlations with band pass times $t_{200-400}$ ... $t_{1400-1600}$ are observed too, however the correlation is weaker in higher frequencies.

Table 1: Correlations in the whole sample (n=230)

Parameters	FET_a	$t_{200-400} – t_{1800-2000}$	$A_{200-400} – A_{1800-2000}$
	r	r	r
FEV_1	-0.54	-0.22 – -0.52	-
FEV_1/FV	-0.84	-0.29 – -0.83	-0.18 – -0.43
MEF_{25}	-0.59	-0.25 – -0.59	-0.14 – -0.25
MEF_{50}	-0.76	-0.28 – -0.73	-0.17 – -0.32
MEF_{75}	-0.81	-0.27 – -0.76	-0.19 – -0.34
$MMEF_{25-7}$	-0.79	-0.28 – -0.74	-0.17 – -0.31
R_{in}	0.50	0.26 – 0.50	0.13 – 0.24
R_{ex}	0.55	0.29 – 0.54	0.14 – 0.26
R_{tot}	0.54	0.27 – 0.53	0.14 – 0.26
SRt	0.58	0.24 – 0.59	0.16 – 0.34
RV	0.46	0.16 – 0.47	0.18 – 0.34
TLC	0.19	0.18 – 0.25	0.19 – 0.43
RV/TLC	0.44	0.18 – 0.43	-
VC	-0.17	-	-
IC	-	-	0.16 – 0.34
ERV	-0.39	-0.18 – -0.37	-

The moderate strength direct correlations of FET_a with bodyplethysmography parameters – aerodynamic tidal expiratory resistance (R_{ex}), specific resistance (SRt), residual volume (RV) and its ratio to total lung capacity (RV/TLC) are found. Moderate correlations of bodyplethysmography parameters with band pass times $t_{200-400}$ ... $t_{1200-1400}$ are observed, which are weaker at the higher frequencies. Band pass energies correlate weaker than band pass times with aerodynamic resistances, RV and do not correlate with RV/TLC at al. Acoustic parameters correlate more strongly with resistances than with volumes, except RV and RV/TLC, reflecting predominantly small airway resistance.

As for correlations in selected groups, FET_a correlates directly with FVC in BA(–) patients, while inverse correlation is seen in patients with obstructive ventilatory defect – BA(+) and COPD (Fig. 1). Correlations of FET_a with FEV_1/FVC and $MMEF_{25-75}$ are unidirectional (inverse) in all groups. The differences in the strength of correlations in groups are insignificant. FET_a inverse correlations with FEV_1 are seen only in BA(+) and COPD groups.

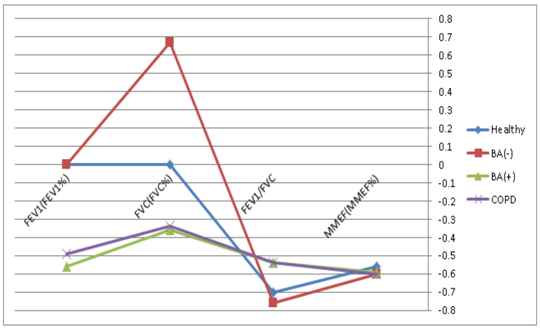

Fig. 1: Correlations of FET_a and spirometry indexes in the groups

There are differences in signs of correlations of FET_a with RV/TLC between BA(+), COPD and healthy, while BA(–) group is indistinguishable from healthy (Fig. 2)

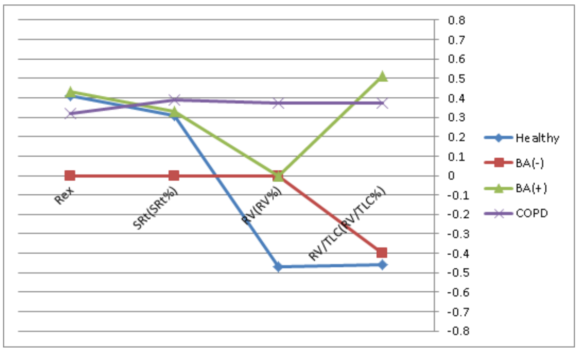

Fig. 2. Correlations of FET_a and bodyplethysmography indexes in the groups

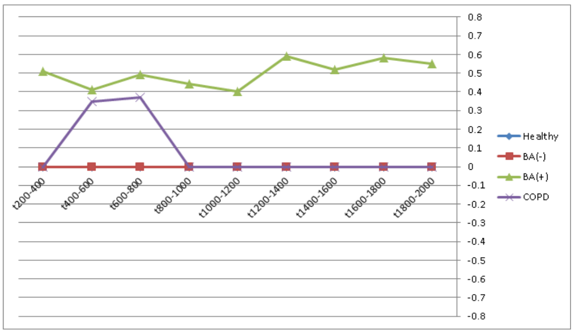

Fig. 3: Correlations of band pass times $t_{200\text{-}400}...t_{1800\text{-}2000}$ with R_{ex} in the groups

It is ironic that R_{ex} correlates with FET_a in healthy individuals, but there is no any correlation in BA(–) group. Supposedly local obstruction in these patients due to its contribution to the increase of R_{ex} retorts existing relationship observed in healthy. There are specific features in correlation of R_{ex} with band times and energies (Fig. 3, 4). There is a correlation of band pass times with R_{ex} only for patients with bronchial obstruction. The difference between BA(+) and COPD in bandwidth of significant correlation is seen. Really R_{ex} in BA(+) correlates with band pass times $t_{200\text{-}400}$ ... $t_{1800\text{-}2000}$, while in COPD it correlates only with $t_{400\text{-}600}$ ... $t_{600\text{-}800}$. There are bidirectional correlations of band pass energies with R_{ex}. The correlation is direct for BA(+), and it is inverse for COPD groups.

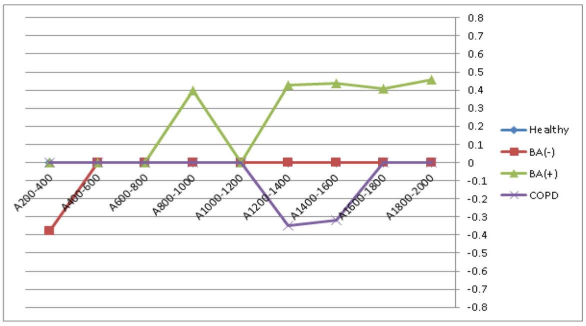

Fig. 4: Correlations of band pass energies $A_{200\text{-}400}...A_{1800\text{-}2000}$ with R_{ex} in the groups

Since the resistance increases in BA(+) and COPD patients by definition, visible differences between groups apparently may be attributed to various origin of R_{ex} growth. Really in BA an increase of resistance happens mainly in large bronchi, which leads to an amplification of linear flow velocity and subsequent elevation of high frequency noise energy due to Strouhal law. Whereas in COPD an increase in R_{ex} is caused by collapse of small bronchi, leading to a reduction of the flow in downstream tree branches and subsequent decrease of high frequency noise energy. However, this effect is seen only for the high frequency range $A_{1200\text{-}1400}...A_{1400\text{-}1600}$ (Fig. 4).

IV. Conclusions

Forced expiratory acoustic parameters in the whole sample of patients and healthy correlate more strongly with resistances and velocities than with volumes, except for residual volume and its ratio to total lung capacity.

Revealed correlations of opposite direction between acoustic parameters and bodyplethismography / spirometry indices in groups of asthma, COPD patients and healthy may be useful for further development of the acoustic diagnostic method.

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Tracheal Sounds of Forced Expiration During Short- and Long-term Immersion

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Abstract— **Experimental data on tracheal sounds duration (Ta) of forced expiration (FE) under immersion conditions are presented. The measurements were conducted in healthy volunteers during short-term water immersion (13 persons) and 5-day dry immersion (7 persons) [1, 2]. In both cases the immersion did not lead to significant changes of group average Ta except for the first day after long-term immersion exposure when it increased significantly. In some people both immersion types caused a significant increase or decrease of Ta. Observed increase was likely the result of intense wheezes occurred at the end of FE. The causes and potential diagnostic value of such wheezes are not clear.**

Keywords— **forced expiration, tracheal sounds, wheezes, immersion, modeling microgravity**

I. Introduction

The increased values of Ta measured during FE maneuver may indicate an obstructive change in respiratory system [3]. Due to its simplicity Ta measurement technique can be used under extreme conditions, e.g. in astronauts.

Immersion is a well-known model used to study the effect of microgravity in humans. During dry immersion the subject is putting in thermoneutral water covered with a special elastic waterproof fabric and so remains dry during the study. Dry immersion is also used for medical purposes. The long-term effects of dry immersion may be compared with those of actual space flight. Immersion leads to different physiological changes in pulmonary system caused by the effect of hydrostatic pressure on body surface and especially the chest wall and abdomen. It was established that spiromertic parameters FEV1, FVC, PEV were reduced in the first hours after water immersion exposure as well as in the beginning of space flight [1, 4, 5]; water immersion was shown to alter lung function and respiratory muscle strength [6].

The aim of this study was to evaluate the potential of the acoustic technique for monitoring changes in respiratory function during immersion with particular attention paid to the individual response to the exposure.

II. Methods

Tracheal sounds were recorded during FE by a small accelerometer placed on the side of the neck over the larynx. Subject while wearing the nose clip held the sensor by hand and performed FE standard maneuver. All subjects breathed through a tube of flow sensor because spirometric data were collected simultaneously. Subjects performed 5-10 FE maneuvers in all series. The signals were recorded and analyzed with SpectraLab (Sound Technology Inc, USA) in the frequency range 200 Hz – 2 kHz. All records were tapped to eliminate signals containing the unwanted noise or cough. Ta was evaluated by expert listening as well as on the basis of signal intensity drop to the threshold level of -60 dB corresponding to the background noise level. Statistical analysis was performed using Friedman and Wilcoxon signed-rank tests. Differences were considered statistically significant at $p<0.05$.

Short-term water immersion. The study involved 13 healthy volunteers aged 20 to 50: 12 men and one woman. The subjects were immersed in water at the level of the clavicles and were half-upright. Acoustic measurements were conducted after 1-1.5 hours after immersion. Control measurements were performed in a sitting position under normal conditions.

Dry 5-days immersion. The study involved 7 healthy men aged 21 to 25 years. Acoustic examination included 4 series: before the immersion, on the first and fourth day of immersion and the next day after. During control and immersion studies subjects were in a prone position.

III. Results

FE tracheal sound signals. The signals are a broadband noise with a number of narrow spectral peaks and harmonics. To calculate Ta we should know the time of the beginning and of the end of exhalation. The starting time of the correctly executed FE can be readily found either manually or automatically. The end can be easily determined in case of smooth decrease in signal intensity over time but it's not the only possible pattern. Many subjects demonstrate intense wheezes during FE and their acoustic signals have a complex shape so Ta value becomes ambiguous (Fig.1). Wheezes are tonal sounds of different frequencies. The well-audible wheezing may continue at minimal respiratory flow when the broadband noise amplitude is falling up to background level.

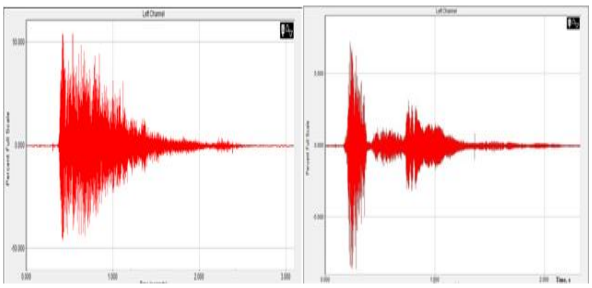

Fig.1: Time series of FE tracheal sounds. Left: broadband noise with slight wheezes. Right: intensive wheezes at the end of FE

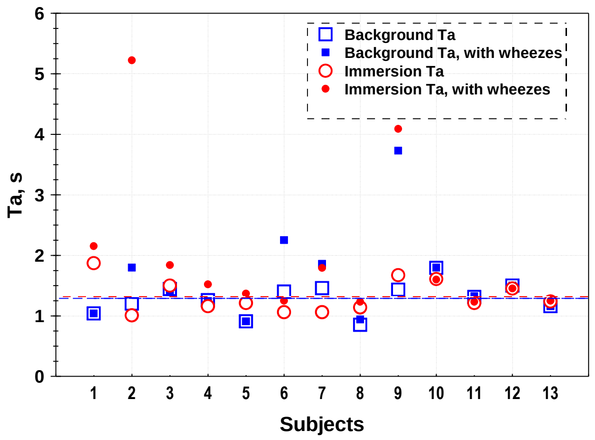

Fig.2: Individual dynamics of Ta during water immersion. The numbers on the X-axis correspond to different subjects. Big square and big round markers show the background and water immersion Ta values, respectively (in seconds). Small square and small round markers show the same Ta values but including the duration of wheezes. Two dashed and almost identical lines indicate group average background and immersion Ta.

In this study Ta was determined as the moment of falling FE broadband noise intensity to the level of background noise but an estimate of Ta taking into account the duration of wheezes was also performed. The results obtained for water immersion are shown in Fig.2.

Effect of short-term water immersion. The group average background Ta was equal to 1.29 ± 0.26 s and the individual background Ta ranged from 0.9 to 1.8 s. During water immersion the average Ta was 1.32 ± 0.27 s. The immersion effect has not led to significant changes in average Ta (p = 0,31).

Effect of 5-day dry immersion. The group average background Ta was equal to 1,35 ± 0,42 s and individual baseline values ranged from 0.8 to 2.1 s.

The average Ta in the first day of dry immersion was 1.33 ± 0.37 s, on the fourth day 1.35 ± 0.67 s, and on the first day after immersion Ta became 1.60 ± 0.57 s. Statistically significant growth was obtained only for the period of adaptation and accounted for 17% comparing with background value (p=0, 018). Individual values in the recovery period ranged from 1.1 to 2.6 s.

Individual dynamics. During water immersion the individual increase and decrease of Ta up to 30% was observed. Two subjects demonstrated especially conspicuous wheezes in the end of FE. One of them had wheezes in normal conditions as well as during immersion while another wheezes appeared during the exposure and his Ta with wheezes exceeded 5 s. According to our observations intensive wheezes often appear with age; both subjects were actually older than others. During 5-day dry immersion the pronounced dynamics different from all others participants was observed in one subject. On the 4 day of immersion his Ta increased by 33% compared to the baseline up to 2.8 s and remained increased (2.6 s) after exposure. This man showed up the intensive wheezes in all experimental series including normal conditions.

IV. Discussion

In our study the short-term water and 5-day dry immersion did not cause any changes in group average Ta. However some subjects demonstrated expressed multidirectional changes. Despite the small number of observations these findings are very important and stress the role of individual reactions during respiratory system adaptation to the immersion and, apparently, microgravity. The significant increase in Ta after 5-day immersion is also of great interest especially because spirometry data at the same time were within normal range [2]. Probably it indicates a high sensitivity of the acoustic technique. These changes may point on an increase in airway resistance due to the body fluid redistribution but this assumption needs to be tested. Another possible reason may be a change in the respiratory muscles coordination found in the first day after spaceflight [7].

There are at least two possible mechanisms of Ta change in immersion: the compression of lungs under water pressure that increases bronchial resistance and should increase Ta and the decrease in lung volumes in the initial period of immersion and microgravity [1, 2, 4, 5, 6] that should reduce it. The overwhelming influence of one of these factors will cause an appropriate change of Ta. The individual lung structure also can make a difference.

The nature of FE wheezes are understudied. The wheezes at the end of FE complicate Ta definition and are highly variable even in one person. This fact should be took into account in the further development of the acoustic method. Ta determined on the basis of the broadband noise intensity fall was proved to be a quite stable acoustic parameter.

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Breath Sounds as Biomarker of Bronchial Asthma

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Abstract— **We analyzed breath sounds in thirty-six rather stable asthmatics and examined their pulmonary function. As a parameter of breath sounds we used power ratio of expiratory-inspiratory breath sounds in their low frequency (100-195Hz) range (E/I L). The E-I L correlated well with the pulmonary function. The E-I L by lung sound analysis correlated well with pulmonary function in patients with asymptomatic asthma. These findings suggest that breath sound can be a useful biomarker to assess airway obstruction and then possibly airway inflammation in rather stable asthmatic patients.**

Keywords— **asthma, breath sound, airflow obstruction, asthma monitoring, biomarker**

I. Introduction

The broncho-provocation studies suggested that an increase of pitch or intensity of lung sounds was a common and early finding of airway narrowing. [1, 2] The changes in these sound parameters correlated well with forced expiratory parameters in lung function tests and may reflect pathological changes in the airway. We reported that breath sounds was useful in the monitoring of rather stable childhood bronchial asthma. [3, 4] To clarify this in adult asthmatics, we analyzed lung sounds of asymptomatic adult asthmatic patients and compared the lung sound parameters with their pulmonary function.

II. Subjects and Methods

Subjects: Thirty-six subjects with stable bronchial asthma (Age 41 ± 13.2, M/F: 12/24, FEV1/FVC%: 75 ± ±9.9) were examined while they were free from asthmatic symptoms. This study protocol was approved by our local ethics committee.

Methods: Lung sounds were recorded for 30 seconds over the base of the left lung by using a hand-held microphone. The sound recording was performed in a quiet room while the patients breathed freely through a mouthpiece of heat sensing pneumo-tachograph. The recording system consisted of a microphone (Bio-Sound Sensor BSS-01; Kenz Medico, Saitama, Japan), a signal processing system, and a personal computer. The sensor had a band-pass filter range of 40 to 2500 Hz and sound-collecting ability in the 40 to 2000 Hz range. The recorded sounds were analyzed by using a sound spectrometer (LSA-2008; Kenz Medico, Saitama, Japan). The recorded sounds were resampled to 10,000 Hz and analyzed by 1024-point fast Fourier transformation with 60% overlap into adjacent segments by using a Hanning data window. The sound intensity (dBm) was used to express the power of the sound in this study we compared expiratory-inspiratory ratios of sound power (dBm) in the low-frequency (100-195Hz) range (L E-I) with pulmonary function parameters.

III. Results

Our new lung sound parameter L E-I correlated well with FEV1/FVC % (Fig. 1), FEV1/Predicted FEV1 % (Fig. 2), V50/ Predicted V50% (Fig. 3) and V25/ Predicted V25% (Fig. 4).

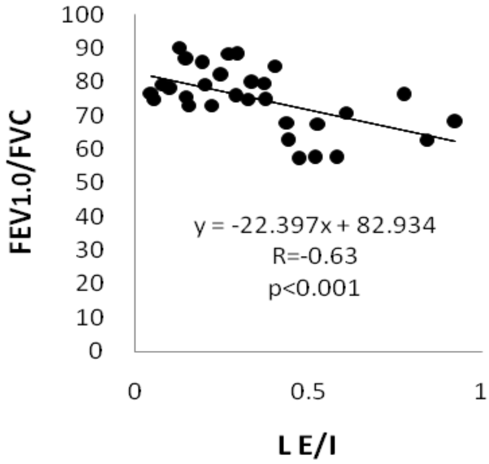

Fig. 1: Correlation of LE/I with FEV1/FVC%

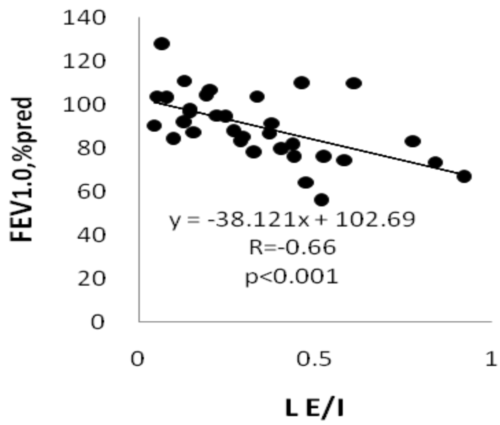

Fig. 2: Correlation of LE/I with FEV1/ Predicted FEV1%

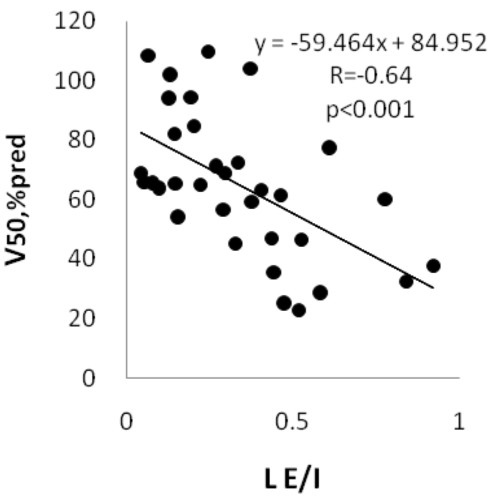

Fig. 3: Correlation of LE/I with V50/ Predicted V50%

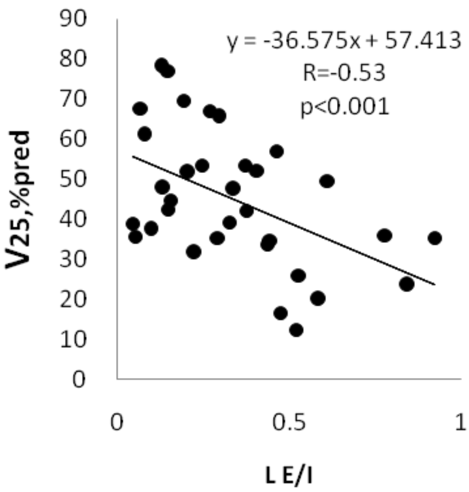

Fig. 4: Correlation of LE/I with V25/ Predicted V25%

IV. Discussions

In our management of bronchial asthma, auscultation of the lung gives us important information. One of the most important information is on the airway inflammation. When there is airway inflammation, airway narrows and the airflow through the airway increase the velocity and produce bronchial breath sound in the peripheral airways. When we listen bronchial breath sound, i.e., can listen expiratory breath sounds clearly in the chest apart from the central airways, we can recognize that the control of asthma of that patient is less than optimal. This means that power ratio of expiratory/inspiratory breath sound is a good indicator of airway narrowing in asthmatic patients.

In this study we showed that the L E-I correlated well with pulmonary function in asthmatic patients. We selected the low frequency range betgween100 and 195 Hz to represent intensity of expiratory/inspiratory breath sound. The power of recorded sound below 100Hz is often contaminated by environmental noise. The power of higher frequency range is weaker than low frequency range and may be contaminated by friction noise. We could find good correlation between L E-I and parameters of airway obstruction.

These findings in this study suggest that our clinical auscultatory findings represent airway obstruction of mild degree in rather stable adult asthma patients.

V. Conclusions

The E-I LF by lung sound analysis correlated well with pulmonary function in rather stable asthmatic patients. As pulmonary function are known to reflect airway inflammation in asthma, these findings suggests that breathe sound can be a useful biomarker to assess airway inflammation in rather stable asthmatic patients.

Acknowledgment

We thank organizing committee of ILSA 2015 for their kind support in this presentation.

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Development of a Digital Filter to Convert Mic-Sounds to Stethoscope-Sounds

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Abstract— **It is possible that the lung sounds on auscultation using a stethoscope were quite different from those acquired by an air-coupled microphone. We examined the transfer function between the mic-sounds and stethoscope-sounds on the neck over the trachea, where white noise was introduced from the mouth. The frequency response was not flat, with marked difference among stethoscopes according to the brand. We developed digital filters to convert mic-sounds to sounds similar to stethoscope-sounds using the transfer function. The performance of the filters was tested for a pulse wave on the neck and normal breath sound on the chest wall, demonstrating excellent agreement between the filtered mic-sounds and stethoscope-sounds. It was shown that time expanded waveform of fine crackles may be noticeably changed by these filters in terms of time-axis measurements. Moreover, the filters seem useful to produce educational materials about auscultation.**

Keywords— **Stethoscope, Air-coupled microphone, Transfer function, Crackles**

I. Introduction

Lung sound analysis is usually performed for sounds acquired from a microphone on the chest wall. However, we listen to lung sounds from a stethoscope in clinical practice. Therefore, it is possible that there are discrepancies between the result of lung sounds study and our impression in clinical practice. To overcome this drawback, we analyzed the acoustic characteristics of stethoscopes on the human body surface and we made a digital filter to convert sounds from a microphone (mic-sound) on the chest wall to sounds similar to those heard by stethoscope (stethoscope-sound).

II. Methods

Transfer function of stethoscopes: Sounds from a stethoscope (only diaphragm type was tested) and an air-coupled microphone, which were placed on the neck over the trachea (right and left, respectively), were simultaneously recorded on an IC-recorder, while white noise generated by a loud speaker was introduced into the mouth (Fig. 1). The recorded data were analyzed on a personal computer; coherence and transfer function between both channels.

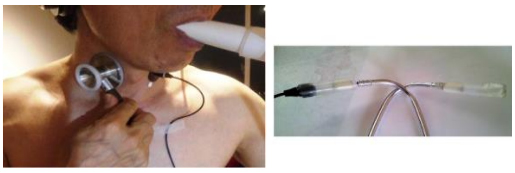

Fig. 1 Lt: Stethoscope and air-coupled microphone on the neck. Rt: Ends of the stethoscope; one end is connected to a microphone and the other end is occluded by a silicon tube.

Digital filter to convert the mic-sounds into the stethoscope sounds: The impulse response between both channels was calculated from the transfer function using an inverse-FFT. Convolution using this impulse response was used as the digital filtering.

Validation of the digital filtering: The digital filtering method was validated using following sounds samples.

1. A repetitive pulse wave was introduced into the mouth and sound recording on the lateral neck was performed bilaterally using a stethoscope and an air-coupled microphone.

2. Normal breath sounds on the chest wall were recoded using the air-coupled microphone and stethoscope, which were placed on the chest wall closely to each other.

III. Results

Transfer function of the stethoscopes: The coherence between signals from both channels was nearly 1.0 within a frequency range of 50 Hz to 1 kHz. Frequency response was markedly different according to the stethoscope brand. The Littmann brand stethoscopes had a prominent tendency of low frequency sounds exaggeration. In contrast, the Kenzmedico brand stethoscopes had relatively flat frequency characteristics at the bandwidth of 50 to 600 Hz.

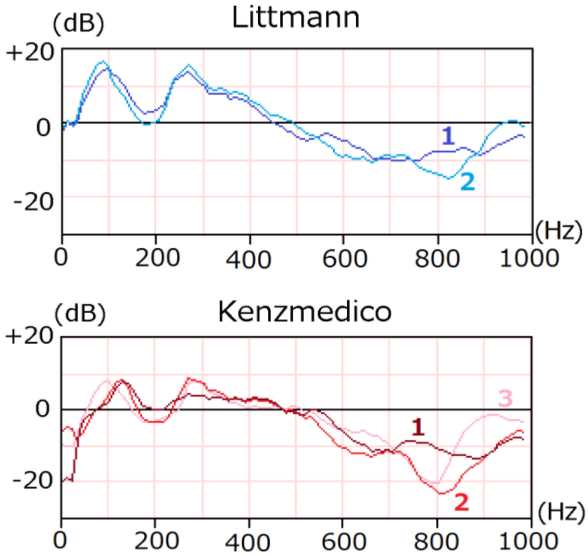

Fig. 2: Frequency response of stethoscopes

Littmann 1: Cardiology III, 2: Classic II. S.E.
Kenzmedico 1: DoctorphonetteNEO, 2: Flairphonette, 3: No.120

Impulse response of the stethoscopes: The impulse response between both channel signals was similar among stethoscopes of a brand but was very different between the two brands.

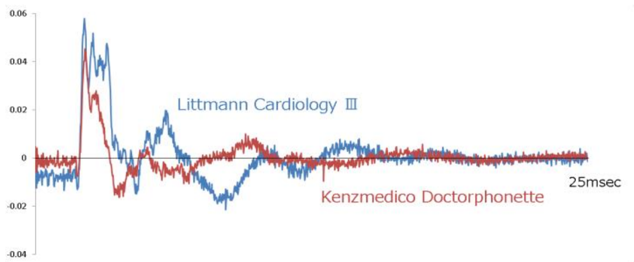

Fig. 3: Impulse response

Validation of the filtering:

1. Pulse wave

Filtered mic-sound using each impulse response was similar to respective stethoscope-sound (Fig. 4).

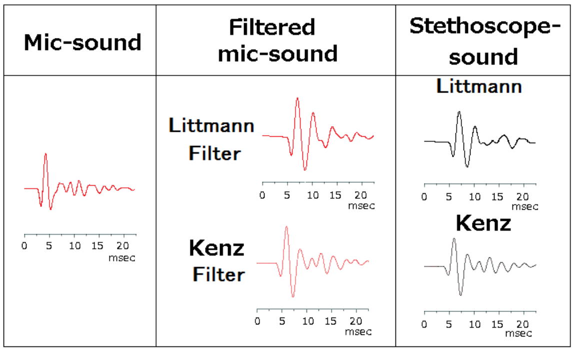

Fig. 4: Pulse wave recorded on the lateral neck surface

2. Normal breath sound

Power spectra of filtered mic-sound using each impulse response were similar to those of respective stethoscope-sound (Fig. 5).

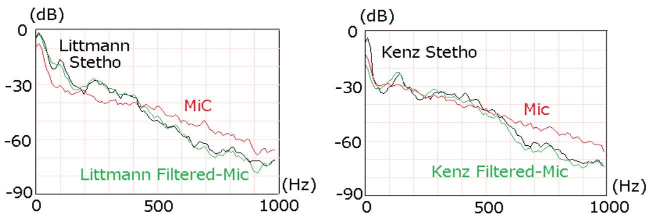

Fig. 5: Power spectra of normal breath sound on the chest wall

IV. Discussion

Stethoscope sounds were quite different from mic-sounds. Surprisingly, the waveform of stethoscope-acquired pulse wave was also different according to the stethoscope brand. We tried to develop stethoscope-specific digital filters to convert mic-sounds to stethoscope-sounds. The filtered mic-sounds were very similar to the respective stethoscope-sounds.

Previous studies about acoustic properties of stethoscopes were usually performed using an artificial coupler with sound source [1-3]. At first, we tried to obtain the transfer function from the mic-sounds and stethoscope-sounds acquired on a rubber plate or Konnyaku (Konjac gel) over a loud speaker. However, the digital filters derived from these performed poorly when applied to lung sounds on the chest wall. Therefore, we decided to calculate the transfer function using the sounds on the neck, from which we could make the digital filters with excellent performance.

The difference of acoustic characteristics among lung sounds sensors may have clinical relevance. Currently, crackles are classified into “fine” and “coarse” based on measurement of the waveform. For example, a fine crackle from an IPF patient had an initial deflection width (IDW) of 0.8 ms in the mic-sound. However, the stethoscope filtering changed the IDW to be 1.2 ms and 1.6 ms, for the Kenzmedico and Littmann stethoscopes, respectively. Indeed, auditory sensation of crackles as well as normal breath sound is quite different according to the brand of stethoscope. The fact means the current definition may not be applied to auscultation sounds.

The present study has a limitation that the air-coupled microphone may not have flat frequency response [4], which should distort the estimated frequency response of stethoscopes. However, the filters to convert mic-sound to stethoscope-sound may be useful, e.g. to produce educational materials about auscultation.

V. Conclusions

Frequency response of stethoscopes was markedly different according to the stethoscope brand. We developed digital filters to convert mic-sounds to sounds similar to stethoscope-sounds of two brands, which were demonstrated to have good performance on the neck and chest wall. Such a method is considered to be useful to bridge the gap between auscultation findings and lung sounds study.

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An Approximate Estimation of Forced Expiratory Bronchial Resistance in Asthma and COPD Patients by Means of Biomechanical and Acoustical Surrogate Measures

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Abstract – **A strong asymmetry in the distribution of ratios of biomechanical and acoustic surrogate measures of bronchial resistance in spirometry positive asthma and COPD is revealed, which presumably reflects the heterogeneity of obstructive diseases in terms of extension of bronchial obstruction and predominant location of obstruction by the levels of bronchial tree.**

Keywords – **forced exhalation, resistance, acoustics, biomechanics, statistical analysis**

I. Introduction

Measurements of a bronchial resistance under forced exhalation (FE) is potentially applicable to detect human bronchial obstruction. However direct measurements of the resistance are complex and require the use of specialized and rare bodyplethismography equipment. We have proposed biomechanical, acoustic-biomechanical and acoustic-anthropometric expressions for an approximate estimate of forced expiratory bronchial resistance which were named surrogate measures [1]. The objective of this study is to test this hypothesis in the independent sample of patients.

II. Materials and methods

The proposed surrogate estimates include biomechanical (measuring maximal expiratory mouth pressure and flow rates of the exhaled air flow) measures the resistance in the middle of FE maneuver $z_2 = P_{emax}/MMEF$, in the end of FE maneuver $z_3 = P_{emax}/MEF_{75}$, where P_{emax}, cm H_2O – maximum expiratory static pressure measured at mouth, MEF_{75}, l/s – maximum expiratory flow rate at the level of 75% forced expiratory volume capacity (FVC, l), MMEF, l/s – medium maximum expiratory flow rate 25% - 75% FVC.

On the other hand, the acoustic-biomechanical measure of forced expiratory bronchial resistance $z_{ab} \sim FET_a P_{emax}/FVC$, as well as the acoustic-anthropometric one $z_{aa} \sim FET_a/H$, are developed, where FET_a, s is the tracheal forced expiratory noise time in frequency band of 200-2000 Hz [2], and H, m is height of body. We found previously [1] in the sample of 91 healthy young males and females (17-25 years) that biomechanical forced expiratory surrogate measures z_2, z_3 and their ratios to acoustic surrogate measures z_{ab}, z_{aa} are in limits of $Me(Q_5;Q_{95})$: z_2 = 19.3(11.8;30.5), z_3=33.3(17.8;58.2), z_2/z_{ab} = 0.56(0.40;0.82), z_3/z_{ab}=0.97(0.67;1.5), z_2/z_{aa} = 23.8(14.3;38.7), z_3/z_{aa} = 40.5(24.6;67.3).

In the work we studied the biomechanical surrogate measures of bronchial resistance and ratios of these parameters to acoustic surrogate measures in the independent sample of patients (age 16 – 68 years). The sample contains 24 spirometry negative asthma patients – BA(-), 24 spirometry positive asthma patients – BA(+), 25 COPD patients. Forced expiratory tracheal noises recording was made in accordance with previously described methodology [2]. The maximal FET_a value in 3 well done attempts was used for subsequent analysis. Mechanical parameters of lung function were determined by spirometry and bodyplethysmography with diagnostic complex MasterScreen Body in accordance with ATS/ERS 2005 guidance. Spirometry indexes FEV_1, FVC, FEV_1/FVC were assessed. Body plethysmography parameters bronchial resistance of tidal expiration R_{ex}, functional residual capacity FRC, residual volume RV, total lung capacity TLC and their ratio RV/TLC were evaluated. Maximum static expiratory mouth pressure P_{emax} (cm H_2O) was measured with Micro RPM (Micro Medical Ltd.). All subjects gave their informed consent.

III. Results and discussion

The diagrams "box and whiskers" for ratios of surrogate resistances z_2/z_{ab}, z_3/z_{ab}, z_2/z_{aa}, z_3/z_{aa} in studied groups are built (see Fig. 1, for example).

The values of the coefficients of skewness of all analyzed parameters in studied groups of patients with obstructive lung diseases are listed in Tab. 1. The values of the coefficient of skewness which are greater than 1.0, characterizing highly asymmetrical distribution are typed bold.

A significant gradual increase in z_2, z_3 when moving from BA(-) group to BA(+) group and to COPD group is of interest for bronchial obstruction diagnostics. If we take as a threshold the 95% limit of z_2 = 30.5 (5), an excess of FE bronchial resistance above the norm may be detected in 75% of patients of BA(-) group, and 96% of patients in BA(+) and COPD groups. Similar results are obtained if 95% limit of z_3 = 58.2 (6) is used.

Considering behavior of ratios z_2/z_{ab}, z_3/z_{ab}, z_2/z_{aa} (Fig. 1), z_3/z_{aa}, one can see that, despite the growth of their numerators and denominators in the groups the ratios itself look differently. A significant bias between medians of BA(-) and BA(+) groups is lost. However, according to Table 1 the distributions of these parameters in BA(+) and COPD groups become highly asymmetric (> 1.0). Furthermore a shift of distribution tails to higher ratios values, than in BA (-) group is seen (Fig. 1).

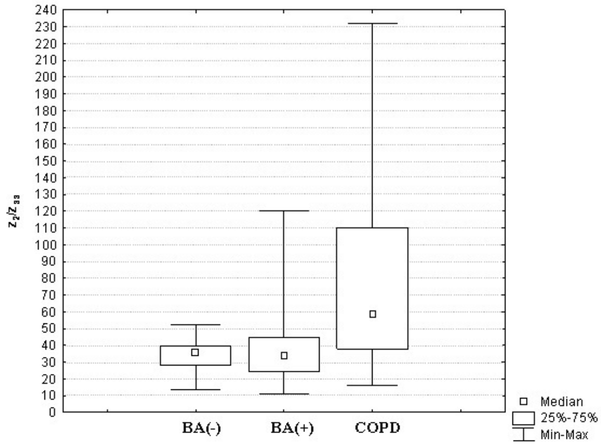

Fig. 1: Box and whiskers diagram of z_2/z_{aa}

Table 1 - The coefficients of skewness of analyzed parameters in groups

Parameter\Gr	BA(-)	BA(+)	COPD
FEV_1/FVC	0.52	0.11	-0.85
FEV_1	0.50	0.16	0.51
MMEF	**1.7**	0.31	0.79
FET_a	**2.07**	**2.23**	0.47
R_{ex}	0.31	0.32	**1.52**
RV/TLC	-0.08	0.13	0.64
z_2	-0.16	0.89	0.18
z_3	0.56	**1.10**	0.26
z_2/z_{ab}	-0.94	**1.37**	**1.28**
z_2/z_{aa}	-0.26	**2.00**	**1.24**
z_3/z_{ab}	-0.24	**3.02**	**2.09**
z_3/z_{aa}	0.08	**3.03**	**2.21**

It is interesting to compare the skewness of analyzed ratios distribution with parameters commonly used for bronchial obstruction diagnosis. As we can see from Table 1, a strong asymmetry in the distribution is characteristic only for R_{ex} in COPD group, whereas in BA(+) group it is observed only for FET_a and z_3.

From acoustic-biomechanical point of view the noise production under FE occurs presumably within the range of the 0th – 7th generations of bronchial tree due to Reynolds numbers (above 2000) sufficient to form turbulent air flow in these airways [3]. In this light, z_2/z_{ab}, z_3/z_{ab}, z_2/z_{aa}, z_3/z_{aa} ratios location within predicted 5% - 95% limits (7 - 10) corresponds to a proportional change in biomechanical and acoustic measures, and can be regarded as a sign of uniform increase of biomechanical and acoustic measures of bronchial resistance.

If actual values of the analyzed ratios are above the 95% predictions (7 - 10), an increase in biomechanical bronchial resistance measures is greater than the increase in acoustic measures, i.e. FET_a is disproportionately reduced. Based on our acoustic model representations, such an effect can be expected under distal displacement of the region of maximum bronchial resistance due to shift of obstruction into smaller than the 7th bronchial tree generations which are "dumb“ in terms of FE noise recorded above trachea.

On the contrary, if actual values of analyzed ratios are lower than 5% predictions (7 - 10), an increase in biomechanical measures of bronchial resistance is less than an increase of acoustic measures, i.e. FET_a is disproportionately high. This effect can be acoustically interpreted as an evidence of the local (regional) obstruction of large bronchi (0th-7th generations).

It is known that COPD is a phenotypically heterogeneous disease – emphysematous and bronchitis types [4]. At the same time asthma is also characterized by heterogeneity (severity, the duration of the disease and so on), which has the structural-morphological basis. This heterogeneity can be interpreted in terms of bronchial obstruction extension (generalized – local), and the predominant level of obstruction in bronchial tree (large – small bronchi). Thus the asymmetry in the distribution of ratios of surrogate biomechanical and acoustic measures of FE bronchial resistance may reflect this phenotypic heterogeneity of BA(+) and COPD groups. Consequently an assessment of the ratios of biomechanical and acoustic surrogate measures of bronchial resistance may be useful to find differences between nosology forms of obstructive pulmonary diseases (asthma and COPD), as well as between phenotypes within these nosologies. A direct comparison of the findings with CT, MRI imaging and other sensitive techniques is welcome.

IV. Conclusions

An excess of the value of biomechanical surrogate measures of bronchial resistance z_2 и z_3 above the 95% limits of the forecast in healthy [1] is found in the majority of patients with asthma and COPD.

A strong asymmetry in the distribution of ratios of biomechanical and acoustic surrogate measures of bronchial resistance in spirometry positive asthma and COPD is revealed, which presumably reflects the heterogeneity of obstructive diseases in terms of extension of bronchial obstruction and predominant location of obstruction by the levels of bronchial tree.

The study was supported by RFBR grant 14-04-00048.

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Bronchial Obstruction Patients Clusterization by Means of Two-Dimensional Analysis of Forced Expiratory Tracheal Noise Time and Band Pass Energy

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Abstract – **The clusterization of patients found is caused by differences in the severity of airway obstruction, which from modeling positions may be connected to the difference of the relative contribution of large bronchi and more distal airways in the pulmonary ventilation defect.**

Keywords – **noise, forced exhalation, time, energy, partition, bronchial obstruction, asthma, COPD**

I. Introduction

It has been demonstrated that forced expiratory (FE) noise time analysis is an effective method of diagnosis of airflow obstruction in patients with bronchial asthma (BA). The objective is an investigation of discrimination ability of newly developed acoustic parameters of FE tracheal noises in the sample involving healthy and patients with obstructive lung pathology – asthma (BA) and chronic obstructive pulmonary disease (COPD).

II. Materials and methods

Both gender subjects were studied – healthy nonsmokers (n=50) aged Me-25.0; LQ-21.0;UQ-30.0, patients with obstructive lung diseases – BA (n=73), aged Me-37.0; LQ-21.0; UQ51.0 and COPD (n=35), aged Me-58.0;LQ-54.0;UQ-62.0. BA and COPD diagnosis were identified early by experienced pulmonologist. The subjects were interviewed by a specially developed questionnaire. Spirometry (before and after salbutamol inhalation) and body plethysmography were performed with apparatus Master Screen Body (Erich Jaeger, Germany) according to ATS/ERS 2005 guidelines. Registering and computer processing of tracheal FE noises were carried out. FE acoustic parameters – FE noise time (FET_a) and energies of noise in 200-Hz frequency bands, normalized by common noise energy in the band of 200-2000 Hz, ($A_{200-400}/A_{200-2000}$, $A_{400-600}/A_{200-2000}$, $A_{800-1000}/A_{200-2000}$,…) were assessed by specially developed algorithms. After spirometry assessing only the asthma patients with obstructive ventilation defect (n=41, aged Me-37.0, LQ-26.0, UQ-54.0) have been selected for following analysis.

III. Results and discussion

An application of the combination of FET_a > 2.26 s "or" $A_{800-1000}/A_{200-2000}$ > 0.117 criteria (Fig. 1) provided a distinguishing healthy and obstructed patients (BA and COPD) with sensitivity 88% and specificity 90%.

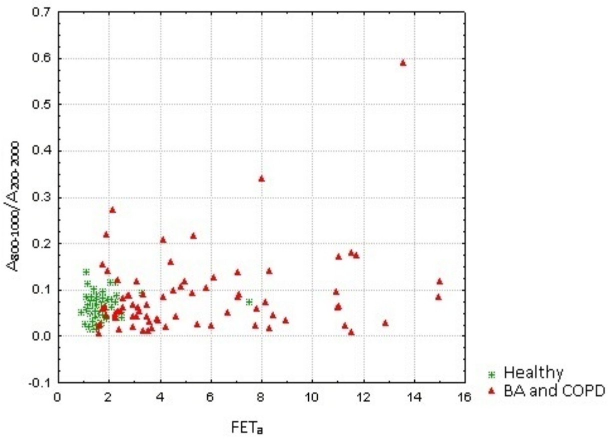

Fig. 1: Scatterplots of $A_{800-1000}/A_{200-2000}$ ratio and FET_a in healthy and BA, COPD patients

Studying other combinations of acoustic parameters we found that in the 2-dimentional feature space – FET_a, $A_{400-600}/A_{200-2000}$ – the sample of patients with asthma and COPD is divided into two distinct "branches" (Fig. 2).

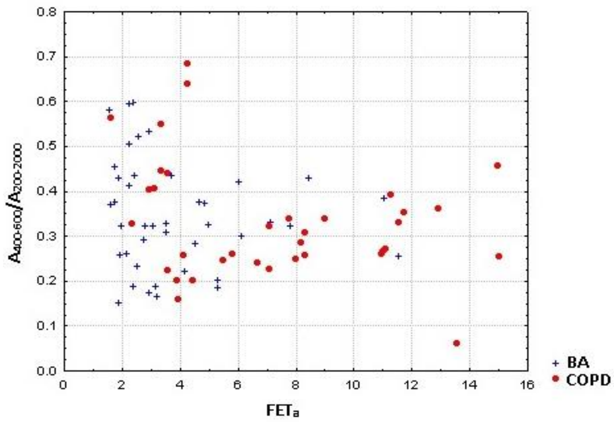

Fig. 2: Scatterplots of $A_{400-600}/A_{200-2000}$ ratio and FET_a in patients

The frequency band of 400-600 Hz is typical for mid-frequency forced expiratory wheezes, which in accordance with [1] commonly are produced in large airways (from trachea to segmental bronchi) by turbulent air flow. It's well known that sound pressure of noises, produced by turbulent flow in tubes with varying cross section, is proportional to a squared linear velocity of the flow [2]. Therefore, from acoustic point of view an elevation of energy of noise production in the band of 400-600 Hz in relation to the total FE noise energy, fixed in the band of 200-2000 Hz (i.e. a rising the ratio $A_{400-1000}/A_{200-2000}$), can be caused by an increase of the linear velocity of the turbulent airflow in large bronchi. The effect can be achieved under large bronchi obstruction in which by reducing the cross section of the air lumen, at the same volume velocity as in norm, the linear flow velocity should increase. Indeed, in many asthmatics with large bronchi obstruction, a more noisy FE is observed than in healthy. On the contrary, within the framework of these ideas, the reduction of ratio $A_{400-1000}/A_{200-2000}$ talks about reducing the linear velocity in the large bronchi, and this may be due to a decrease of air flow volume velocity. The most probable reason for limiting the volume velocity seems to be an elevation of the resistance of more peripheral (upstream) airways. Moreover, the FET_a elongation is also characteristic for this case due to slowing down lung units emptying [1]. Thus, according to the acoustic deductions, left "branch" of the scatterplot Fig. 2 may be associated with a primary obstruction of the large bronchi, while the right – with obstruction of airways, located more distally (i.e. smaller bronchi).

Let's analyze how these acoustic interpretations may be verified by estimates of indicators of lung function.

The linear deciding rule $L = (A_{400-600}/A_{200-2000})/FET_a = 0.075$ was formulated by means of maximum likelihood ratio to separate the “branches” found in the scatterplot (Fig. 2). Thus two clusters L>0.075 and L<0.075 were identified in the sample of patients with bronchial obstruction. A comparison of these clusters by means of nonparametric Mann-Whitney test showed statistically significant distinctions in the most of lung function indices (Tab 1). These distinctions mainly were retained after bronchodilator test (salbutamol).

Table 1: Significance of distinctions of lung function indices in patients between clusters L>0.075, L<0.075 before/after salbutamol inhalation

Groups / Parameters	All patients	BA	COP D
FEV_1, l	0.000009/0.000	0.001/0.0	ns/ns
FEV_1%	0.0002/0.002	ns/ns	0.04/n
FEV_1/FVC	0.000001/0.000	0.003/0.0	0.01/0
R_{tot}	0.001/0.000005	0.002/0.0	ns/ns
R_{tot}% pred.	0.003/0.00001	0.004/0.0	ns/ns
RV, l	0.001/0.003	ns/ns	ns/ns
RV% pred.	0.03/ns	ns/ns	ns/ns
RV/TLC	0.00005/0.0002	0.005/0.0	ns/ns
RV/TLC%	0.02/ns	ns/ns	ns/ns

A severity of bronchial obstruction in respiratory medicine is commonly estimated by FEV_1% of the predicted value. Therefore according to the Table all patients (BA and COPD) included in the cluster L<0.075 have significantly more severe airway obstruction than patients included in the alternative cluster L>0.075. A higher airway resistance R_{tot} in this group of patients and higher values of the indices of lung hyperinflation (air trapping) RV и RV/TLC, reflecting as a part of the whole obstruction of small airways, may be treated in support of this conclusion. It should be noted that patients with L<0.075 are older than ones with L>0.075 (p=0.0003). Nevertheless, there are significant differences in the parameters of pulmonary function, expressed in % of predicted values, which obviously eliminate the effect of age and gender.

We also analyzed whether any particular groups within the selected clusters of asthma and COPD (Tab 1). When splitting the sample by nosology groups became less representative (number of parameters with significant differences is reduced). However, BA patients with L<0.075 are characterized by higher bronchial resistance R_{tot} and increased air trapping index RV/TLC, than BA patients with L>0.075. This effect is not observed in COPD patients. In addition, we compared the parameter L in BA patients with reversible and irreversible (fixed airflow limitation) obstruction. The last group (8/41) is characterized by post bronchodilator FEV_1/FVC ratio reduced below the lower 5% percentile limit of normality, which is considered as a sign of more severe disease. In BA patients with fixed airflow limitation the value of L is smaller than in reversible asthma (p=0.06).

It follows from these observations that the decrease of L below 0.075 is caused by more severe airflow obstruction, often associated with an increase in damage of more distal airways [3]. Thus, the results of analysis of parameters of pulmonary function in the test groups, at least, do not contradict the above developed acoustic representations.

IV. Conclusions

The partition of patients with obstructive lung disease (COPD and asthma) is revealed in 2-dimensional feature space involving tracheal forced expiratory noise time in the frequency band of 200-2000 Hz, and the tracheal forced expiratory noise energy in the frequency band of 400-600 Hz, normalized to the total energy in the band of 200-2000 Hz.

This clusterization of patients is caused by differences in the severity of airway obstruction, which from acoustical and clinical-physiological positions may be connected to the difference of the relative contribution of large bronchi and more distal airways in the pulmonary ventilation defect.

The study was supported by RFBR grant 14-04-00048.

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Application of Polyspectrum Analysis to Diagnostic Signs' Detection of Lung Sounds in Patients with the Chronic Obstructive Pulmonary Disease

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Abstract — **The main goal of the study is the detection of the specific auscultatory diagnostic signs of lung sounds in patients with chronic obstructive pulmonary disease (COPD). The basis of the proposed method is polyspectral analysis. Iterative method has been proposed, which includes several stages of finding diagnostic signs. Bicoherence function is calculated and constructed on first stage, further bifrequencies corresponding with bicoherence coefficient maximum value are analyzed and finally module of skewness coefficient is calculated. Set of all the data makes it possible to infer the presence or absence of pathology in the examined sounds. The proposed method can be used by pulmonologists for additional identification of COPD.**

Keywords — **lung sound, COPD, bicoherence coefficient, skewness coefficient, bifrequency**

I. Introduction

Respiratory system diseases are one of the most common pathologies in the world, leading to the death of patients. COPD is among the five leading death causes in the world. Research has shown that prevalence of COPD is increasing every year [1], so the search for new methods of COPD diagnosis remains an actual problem for scientists. A lot of research work on the identification of COPD, based on various approaches has done in recent years. Among them are measurement of forced oscillations using neural networks [2], an object-oriented methodology [3] processing CT images [4-5] and others.

At the same time electronic auscultation acquires more and more wide application in medicine, which allows to identify and objectivize the specific diagnostic features of lung disease. In this paper, we propose a new method of analyzing breath sounds based on higher order statistics, namely calculation of bicoherence functions and skewness coefficients.

II. Mathematical background

The complex nature of the breath sounds is the cause of the application to their analysis methods of higher order statistics (HOSA). Thus the interest can induce spectral components of the respiratory sounds, as well as phase components. Bicoherence function (BF) is used for such (function asymmetry):

$$\gamma_3(f_1, f_2) = \frac{|B(f_1, f_2)|^2}{P(f_1)P(f_2)P(f_1+f_2)}, \quad (1)$$

where $B(f_1, f_2)$ – bispectrum of a process $\{X(k)\}$ and it is defined as $B_k(f_1, f_2) = X_k(f_1)X_k(f_2)X_k^*(f_1+f_2)$, where $X_k(f_i)$, i =1, 2 is the complex Fourier coefficient of the process $\{X(k)\}$ at frequencies f_i, $X^*(f_i)$ is its complex conjugate; $P(f_i)$, i =1, 2 is the power spectrum at frequencies f_i of the process $P_k(f) = |X_k(f)|^2$.

The bicoherence function is defined by information on the phase structure of the process, herewith the influence of amplitude spectrum structure is eliminate.

The magnitude of BF, $|\gamma_3(f_1, f_2)|$ constitutes a measure of the amount of quadrature phase-coupling that occurs in a signal between any two of its frequency components, due to their nonlinear interactions.

Also the calculation of the skewness coefficients module used for the analysis of respiratory sounds in this study:

$$c_3 = K_3 / \sigma^3 \quad (2)$$

where σ^2 is the variance, K_3 is the third-order cumulants:

$$K_3(\tau_1, \tau_2) = \lim_{T\to\infty} \frac{1}{T}\int_0^T x(t)x(t-\tau_1)x(t-\tau_2)dt \quad (3)$$

Non-zero values of the skewness coefficients allow to evaluate the nature and extent of process deviation from the Gaussian noise within of a one-dimensional distribution.

III. Experimental results

In the current study the respiratory sounds of 83 patients were used. From them 49 persons are sick with the COPD (the II stage in a phase of an exacerbation or a latent exacerbation) and 34 are almost healthy. The lung sounds are registered synchronously by means of four accelerometers attached in any thorax surface points. The state of bronchopulmonary systems of patients was verified previously by standard clinical methods of functional diagnostics, including the X-ray analysis, the spirometry, the determination of diffusive ability of the alveolar-capillary membrane, general clinical studies.

Registered lung sounds of each channel were analyzed separately as independent signals. This approach makes it possible to apply this method to the signals for the single-channel sensors. However, in [6] was considered the possibility of applying the method for the average parameters of 4 channels.

The idea of the proposed method is following.

1) At first maximum value of BF was calculated. It was determined that value γ_{3max} does not exceed 50 for healthy people in 98% of cases. There is a characteristic diagnostic

sign (wheezing, whistle), indicating pathology at values $\gamma_{3max} > 50$ in the signal. For healthy people $\gamma_{3max} < 20$ in 90% of cases. Furthermore, it is proposed to assess visually the three-dimensional representation of BF and its diagonal slice. A distinctive feature of lung sounds BF for healthy patients is that it has, as a rule, a plurality of narrow pointed peaks (Fig.1). The typical form of BF surface of patients with COPD has a wide flat region of high BC values (Fig. 2).

2) If the value γ_{3max} is within the uncertainty, i.e. $20 < \gamma_{3max} < 50$ and if by the BF form is not possible to make a definite conclusion about the phase structure then evaluation of automatically calculated bifrequency pair (f_1, f_2) is assessed. This bifrequency pair presents frequencies on which maximum value of BC is determined. It was found that in uncertainty zone only for patients with COPD there are cases when $f_1 \neq f_2$ and $f_2 \neq 2f_1$. Therefore, cases when $f_1 = f_2$ and when $f_1 = 2f_2$ are not diagnostically valuable because there are almost equal number times in both COPD patients and healthy subjects.

3) Modulus of skewness coefficient c_3 can serve as another additionally specifying characteristic diagnostic sign. It has been determined that module c_3 for healthy patients does not exceed value 0.15 in 82% of cases. For patients with COPD $c_3 > 0.15$ in uncertainty zone is seen in 86% of cases.

As an example Figure 1 and Figure 2 show the typical BF diagonal slices for healthy and patient with COPD respectively. Coefficient γ_{3max} of second channel healthy subject is in uncertainty zone, as well as first channel of patient with COPD. In this case according to the method it is necessary to evaluate bifrequency pair and coefficient c_3 (Table 1 and Table 2). Thus, pulmonologist can be unable to conclude the presence or absence of respiratory artifact indicating pathology.

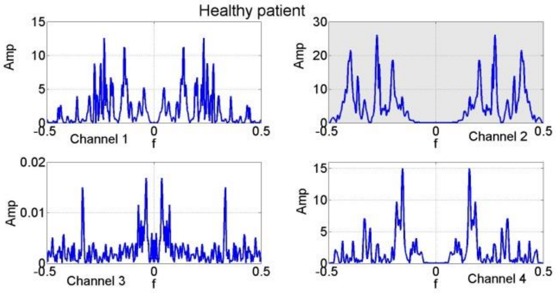

Fig. 1: BF diagonal slices of the healthy subject

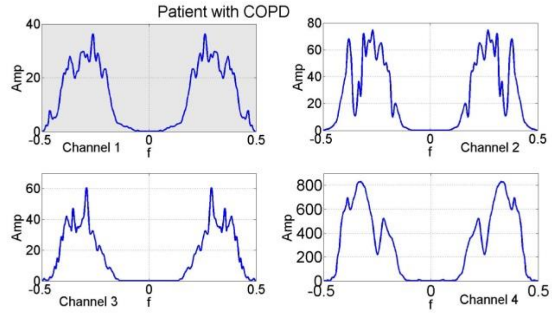

Fig. 2: BF diagonal slices of the patient with COPD

Table 1: BF coefficients γ_{3max}, bifrequency pair and coefficients c_3 of healthy patients

channel No.	γ_{3max}	(f_1, f_2)	c_3
1	12.6	(-0.26563,-0.53126)	0.181
2	26.1	(0.20508,-0.41016)	0.036
3	0.02	(0.037109,-0.074219)	0.047
4	15.0	(0.15625,-0.3125)	0.092

Table 2: BF coefficients γ_{3max}, bifrequency pair and coefficients c_3 of patients with COPD

channel No.	γ_{3max}	(f_1, f_2)	c_3
1	36.3	(-0.26172,-0.47656)	0.160
2	74.6	(-0.26953,-0.46094)	0.054
3	60.8	(-0.29102,-0.41797)	0.181
4	831.2	(-0.32813,-0.32813)	0.905

IV. Conclusions

In this paper the lung sounds analysis of healthy people and patients with COPD was proposed. The analysis method is based on polyspectral analysis of sound signals. As a result of analysis of three calculated parameters (BF coefficients γ_{3max}, bifrequency pair and coefficients c_3) and visual assessment of BF the sounds classification by category "healthy" and "COPD" is made. This approach increases the efficiency of noninvasive diagnostics of COPD and finally allows to select the optimal patients treatment.

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Heart Sound Cancellation from Lung Sound Recordings Using Empirical Mode Decomposition Technique

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Abstract— **In this paper, a new hybrid method for separation of heart sound signals from respiratory sound signals is proposed. It is based on *empirical mode decomposition* (EMD) technique and suppression of noise in signal using *spectral subtraction technique.* The mixed signal is splitted into several components: breathing sounds, cardial sounds and ambient noise.**

Keywords—**heart sound, lung sound, empirical mode decomposition, spectrogram.**

I. Introduction

The human body auscultation is a noninvasive, low-cost and accurate method for assessing heart and respiratory diseases. However, the effectiveness of classic auscultation has decreased due to its inherent restrictions: physical limitations of human ear and the subjectivity of an examiner. This technique got the second breath when it became possible to use electronic devices and computer complexes. However there are several non-trivial technical and theoretical problems on this way. One of them is the problem of separation of heart sound signal from respiratory sounds and ambient noise in combined signal registered on the surface of the chest wall.

Several different techniques have been implemented to reduce the level of heart sound signals within total sound recording. High pass filtering, wavelet based methods, adaptive filtering techniques, fourth-order statistics. Promising results can be achieved by cutting out heart sound segments and interpolating the missing data. Their advantages and disadvantages are analyzed in [1].

In this paper a new hybrid method is implemented, which is based on empirical mode decomposition (EMD), that was recently developed by Norden E. Huang [2] and suppression of noise in signal using spectral subtraction technique [3] for heart sound cancellation in auscultative signals. The mixed signal is splitted into several components: breathing sounds, cardial sounds and ambient noise.

II. Empirical mode decomposition

An EMD algorithm decomposes the signal into a set of functions defined by the signal itself, named the *intrinsic mode functions* (IMFs). When the decomposition level is increased, the complexity of the IMFs decreases, and so does the scale of the signal. It is believed that IMFs preserve the inherent properties of the original signal. Therefore, the EMD can be used to gain significant information inherent to the signal. The EMD produces a bank of IMFs whose sum yields the original signal.

The EMD method decomposes the signal $x(t)$ into IMF $c_i(t)$, i = 1, 2,…, n and residue $r(t)$: $x(t)=\sum_{i=1}^{nn} c_i(t)+r(t)$, where n means the number of IMF. Residue $r(t)$ reflects the average trend of a signal $x(t)$ or a constant value.

Intrinsic mode functions have following characteristics:

- the number of extremes (minima and maxima) and the number of zero-crossings must either equal or must differ by a maximum of one;
- each point, that is defined as mean value of envelopes defined by local maxima and local minima is zero.

The algorithm for searching of intrinsic mode functions is based on a iterative procedure called "sifting". A complete sifting process stops when the residue, $r(t)$ becomes a monotonic function or a function with only one extremum from which no more IMF can be extracted. This decomposinion's method is empirical, intuitive, direct, and adaptive, without pre-determined basis functions. One of the objective of the EMD is to empirically separate a signal into several subsignals of varying, and possibly overlapping, frequency content.

Unfortunately, direct using of EMD to auscultative signal can lead to unsatisfactory results. When the heart sound signal is an object of interest, the researcher has to distinguish it in unevenly distributed noise with large amplitudes. This can lead to a mode-mixing problem, when the same IMFs contain different components of signal and vice versa. Thus a signal can not be clearly separated by merely grouping modes.

For these reasons, in this work it is suggested to use the two-dimensional version of EMD that is applied not to the signal itself, but to its spectrogram. Then the original problem of distinguishing the heart sound signal from breath sounds can be solved by using the method of spectral subtraction.

III. Testing of the algorithm and conclusions

The performance of the proposed algorithm was tested on the real audio signals that were recorded on the chest under different regimes of breathing (0.5 l/min, 1 l/min). As the signal of interest was the heart sound, the breathing sounds were considered as noise.

Evaluation of the quality of separation of signal into components was performed using vectors divergence angle

φ=arcos(cos($y(t)$, $z(t)$)). That is an angle between a vector of signal $y(t)$ and vector of signal $z(t)$. The values of φ change from 0^o to 90^o when functions change from coincidence to full orthogonality (zero correlation).

Since in real-life case a real signal was unknown, the quantitative measure of the effectiveness of the algorithm was the angle between the separated signals, i.e. between the estimate and the residue. The heart sounds and noise (breath+background noises) are assumed to be uncorrelated, therefore the angle between them should be close to 90^o. In the present case $\varphi \approx 79^o$. This result was confirmed by subjective listening and visual comparison of spectrogram

and temporal realizations of the analyzed signal showed in figures 1-2.

Based on the conducted research, it is possible to conclude that the proposed method can be useful in acoustic computer diagnostics of diseases of the cardiovascular and respiratory systems.

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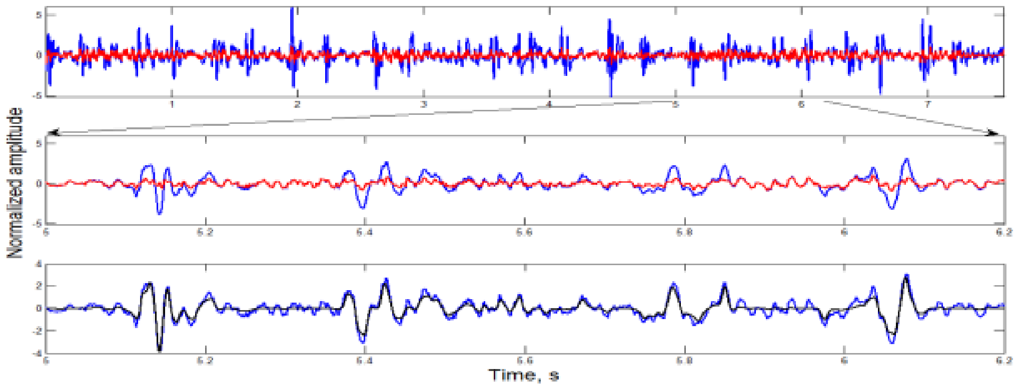

Fig. 1: Separated respiratory and heart sound for real signal: (top panel) blue line – original signal with noise; red line – respiratory signal; (medium panel) blue line – original signal with noise; red line – respiratory signal; (bottom panel) blue line – original signal, black line – extacted heart sound. Amplitude is normalized to rms of original signal

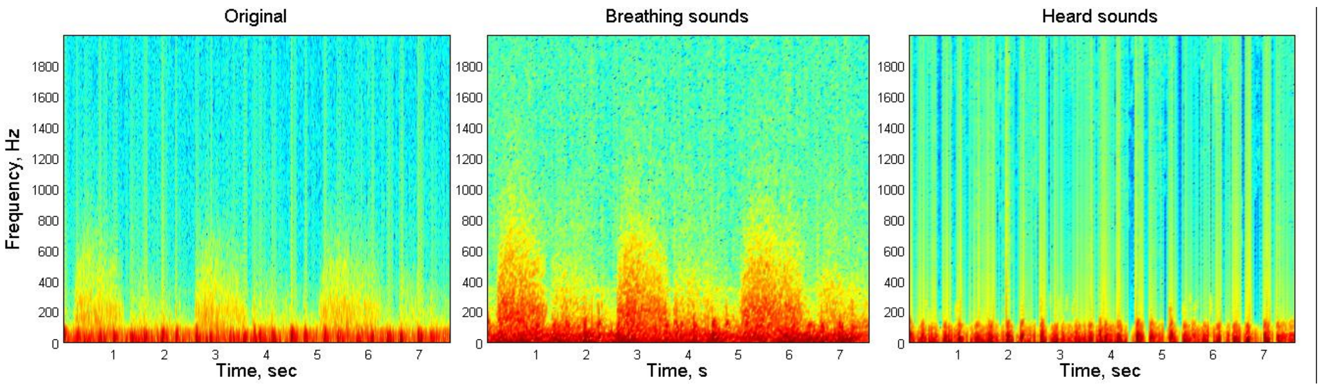

Fig. 2: Spectrograms of the real noised signal (left panel), respiratory signal (medium panel), reconstructed heart sound signal. (right panel)

Bronchodilator Response of Peak Frequency of Forced Expiratory Wheezes in Healthy and Patients with Bronchial Obstruction

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Abstract— **The experimental evaluation of the bronchodilator response of peak frequency (PF) of the most powerful tracheal forced expiratory wheezes (FEWs) and expiratory flows (EF) in representative samples of healthy and patients with bronchial obstruction was carried out. A comparison of bronchodilator responses of PF and EF in healthy and patients made possible to conclude that the most probable mechanism of formation of mid frequency (MF) and early high frequency (HF) FEWs is flow induced oscillations. Whereas the formation of the late HF FEWs is associated with flow independent self-oscillation mechanisms – dynamic flutter in healthy and oscillations of bronchial wall closings in patients with bronchial obstruction.**

Keywords— **forced exhalation, wheezes, peak frequency, expiratory flows, statistical analysis**

I. Introduction

The problem of finding reliable acoustic features of bronchial obstruction remains relevant. Testing under forced exhalation (FE) is used as one of the methods of detecting abnormalities in the function of conducting airways. FE is accompanied by wheezing sounds (FEWs). These sounds are investigated as potential indicators of obstructive lung diseases, but the mechanisms of their formation, as well as localization are still not clear.

There are hypotheses about the FEWs formation by the mechanisms of dynamic flutter and vortex shedding [1]. The model of self-oscillation excitation on closings of bronchial walls at the end of FE maneuver was supposed for patients with bronchial obstruction [2].

The objective is to study bronchodilator response of wheezes peak frequency in order to refine the models of FEWs formation.

II. Materials and methods

We investigated the bronchodilator response of peak frequency of the most powerful mid frequency 400-600 Hz wheezes (MF FEWs) and high-frequency wheezes > 600 Hz (HF FEWs, early and late) recorded above trachea and volumetric expiratory flows (EF).

The first part of the study was carried out with the training sample of subjects, including 71 healthy volunteers and 69 patients with bronchial asthma (BA). The second part of the study was carried out with the control sample consisting of 38 healthy, 36 BA patients and 38 patients with chronic obstructive pulmonary disease (COPD). In all groups, measurements were made before and 15 minutes after inhalation of the bronchodilator (salbutamol).

Peak frequencies (PF) were evaluated in the FE tracheal noise spectrogram by setting the cursor to the fixed time sections t_1=0.1FET$_a$, t_2=0.25FET$_a$, t_3=0.5FET$_a$, t_4=0.75FET$_a$ and the maximum of amplitude of a FEW pass (> 35 dB from peak amplitude level), where FET$_a$ – is a forced expiratory time measured in frequency band of 200 – 2000 Hz.

A spirometry was used to measure flow and volume parameters.

III. Results and discussion

There are statistically significant responses of PF for MF FEWs and early HF FEWs at t_2 time, and at t_3 time for late HF FEWs in healthy of the training sample. Statistically significant response of PF for MF FEWs is observed only at t_4 time in BA group of the training sample. The PF response of the FEWs has the same negative sign in both groups.

As for bronchodilator response of EFs that in healthy having MF FEWs a significant positive response of EFs is observed at time segments t_2 and t_3, while in healthy subjects having early HF FEWs a significant EFs response is seen in all time segments, except t_4. In asthma patients positive response of EFs is seen almost everywhere.

It is interesting that for MF and early HF FEWs in the middle of the FE maneuver EFs have a significant response of about 10% in healthy, and in this case there is, although a small, but significant response of the PFs. In patients the response of the EFs increase essentially, whereas the significant response of the PFs disappear. What does it mean?

Let's consider the model of vortex shedding $f \approx KV(t)/(N\alpha^{3/2}d^3)$, where f – PF, K – Strouhal number (0.2-0.3 or 0.9 for various versions of vortex shedding), V(t) – EF at a time segment t_i, d – bronchus diameter in equilibrium state, α – percentage of bronchus air lumen cross-section under dynamic compression accompanied FE in relation to its inspiratory status, N – number of bronchi in the bronchial tree generation. Thus EF is in the numerator while cross-section of the bronchial tree is in the denominator. When the cross-section increases slightly, this influence on PF change of the MF and early HF FEWs. When the cross-section changes more strongly – EF increases too. In this case, apparently, the numerator and denominator changes retort each other, thereby blocking the PF dynamics.

In contrast, the PF bronchodilator response of late HF FEWs indicates the possibility of involvement flow independent mechanisms in the formation of these sounds. They can be self-oscillating dynamic flutter (in healthy) and self-oscillations of bronchial walls closings (in asthma patients).

In order to test the adequacy of the results, similar assessments are carried out with the control sample of healthy subjects and patients with bronchial obstruction (BA and COPD).

Table 1 : Statistical significance of distinctions (p) / relative change of medians (%) for PFs, EFs before and after a bronchodilator test in healthy

	MF FEW		Early HF FEW		Late HF FEW	
	PF	EF	PF	EF	PF	EF
t_1=0.1FET$_a$ (MEF$_{25}$)	0.62 (n_1=36,n_2=38)	0.45	0.47 (n_1=23, n_2=14)	0.61	-	-
t_2=0.25FET$_a$ (MEF$_{50}$)	0.22 (n_1=38, n_2=37)	0.12	0.67 (n_1=22, n_2=19)	0.51	-	-
t_3=0.5FET$_a$ (MEF$_{75}$)	0.046/-8 (n_1=29, n_2=33)	0.19	0.61 (n_1=6, n_2=4)	0.84	0.56 (n_1=23, n_2=13)	0.9
t_4=0.75FET$_a$	0.6 (n_1=19, n_2=13)	-	-	-	0.53 (n_1=18, n_2=16)	-

Notes: n_1 – number of subjects having FEWs before test, n_2 - number of subjects having FEWs after test.

Table 2 : Statistical significance of distinctions (p) / relative change of medians (%) for PFs, EFs before and after a bronchodilator test in BA and COPD

	MF FEW				Early HF FEW				Late HF FEW			
	BA		COPD		BA		COPD		BA		COPD	
	PF	EF	PF	EF	PF	EF	PF	EF	PF	EF	PF	EF
t_1=0.1FE$_a$ (MEF$_{25}$)	0.52 (n_1=30, n_2=28)	0.005/ 17	0.11 (n_1=35, n_2=36)	0.066	0.40 (n_1=14, n_2=10)	0.022/ 26	0.11 (n_1=20, n_2=20)	0.60	-	-	-	-
t_2=0.25FT$_a$ (MEF$_{50}$)	0.81 (n_1=32, n_2=34)	0.001/ 36	0.69 (n_1=33, n_2=31)	0.086	0.13 (n_1=17, n_2=14)	0.029/ 28	0.25 (n_1=24, n_2=19)	0.38	-	-	-	-
t_3=0.5FET$_a$ (MEF$_{75}$)	0.13 (n_1=16, n_2=23)	0.07	0.78 (n_1=20, n_2=20)	0.000/ -75.2	0.28 (n_1=8, n_2=9)	0.002/ -43	0.41 (n_1=12, n_2=13)	0.000/ -85.4	0.37 (n_1=9, n_2=11)	0.007/ -63	0.40 (n_1=15, n_2=14)	0.000/ -74.8
t_4=0.75FET$_a$	0.39 (n_1=6, n_2=6)	-	0.57 (n_1=8, n_2=10)	-	-	-	-	-	0.61 (n_1=11, n_2=14)	-	0.20 (n_1=23, n_2=23	-

Notes: the same as in tab.1.

According to Table 1 a significant PF response is observed only for MF FEWs in the middle of the FE maneuver in healthy subjects of the control sample. This behavior is similar to the training sample.

A significant PF response is absent in patients of both groups of the control sample (Table 2). A significant EF response in asthma group is seen everywhere except time segment t_3. In COPD group a significant EF response is observed in the middle part of FE maneuver (t_3). These results are also in qualitative agreement with findings made in the training sample. There is no difference in behavior of PFs for asthma and COPD patients.

IV. Conclusions

An absence of significant bronchodilator response of PF in the middle part of FE maneuver is characteristic for patients with bronchial obstruction and on the contrary is not typical for healthy. The findings may be useful for bronchial obstruction diagnostics. However there is no difference in bronchodilator response between asthma and COPD.

A comparison of bronchodilator responses of PF and EF in healthy and patients made possible to conclude that the most probable mechanisms of formation of MF and early HF FEWs are flow induced oscillations. Whereas the formation of the late HF FEWs is associated with flow independent self-oscillation mechanisms – self-oscillation dynamic flutter in healthy and oscillations of bronchial wall closings in patients with bronchial obstruction.

The study was supported by RFBR grant 14-04-00048.

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The Features of Sound Propagation through Human Lungs, Revealed by Transmission Sounding with Phase Manipulated Acoustic Signal

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Abstract - **An application of convolution technique to sounding human respiratory system through mouth and supraclavicular chest areas with complex signal having bandwidth of 80-1000 Hz provided revealing multiple sound transmissions to a chest wall and confirmation of air-structural and structural mechanisms of propagation for different parts of the chest. Global symmetry (left-right) of times of arrivals (velocities) was found under sounding. It was shown that averaged group velocity of sound within bronchial tree lumen did not exceed 150-200 m/s, and decreased from about 270 m/s in trachea as moving deeper in bronchial tree. It was confirmed independently that lengths of transmissions pass in air lumen were much more than in lung parenchyma and therefore air transmission mechanism is substantially more air-born, than previously was thought.**

Keywords - **cross-correlation processing, phase manipulated signal, mechanisms, sound propagation, structural, air-born, velocity.**

I. Introduction

To study poorly known mechanisms of sound propagation in lungs we used convolution technique with sounding respiratory system by complex acoustic signal in frequency band of 80–1000 Hz. The technique provides the compression of pulses with time resolution near 1 ms, and allows to resolve sound arrivals with different propagation ways. 16-channel measurement procedure was used **Ошибка! Источник ссылки не найден.**.

II. Results and discussion

On the sample of 15 healthy volunteers times of propagation of the sounding signal were studied for sensors located in the sounding tube and above trachea (Fig. 1).The first maximum of the envelope in the tube (blue) is seen ahead the first maximum at the trachea (green), i.e. there is a directly propagating sound wave.

The speed of the sound wave in a tract of oral cavity, pharynx, and upper half of trachea estimated by the difference between first maxima is 272±57 m/s for men, which is less than the sound speed in air 354 m/s (37°C).

On the contrary the second arrival is seen ahead at the trachea than in tube (Fig. 1). It means that there is another sound wave traveling in opposite direction. This wave should be produced by a reflection of sounding wave. For the average sound speed of 150-200 m/s we assessed a potential length of wave travel through the lumen of the bronchial tree as 18-23 cm, corresponding to the 11th-17th generations of bronchial tree.These levels of bronchial tree are treated as the reflection area.

And "vice versa" on the basis of the mean anatomical dimensions of bronchial tree length we found that the average group sound speed in the lumen of bronchial tree did not exceed 150-200 m/s.

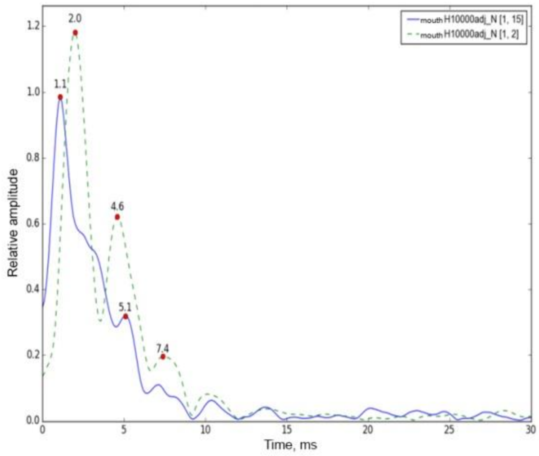

Fig. Ошибка! Источник ссылки не найден.: Convolution curves of the signals registered in the sounding tube and above trachea

Since in the lumen of upper part of trachea sound speed is about 270 m/s the speed of sound wave should decrease when it moves deeper into the lumen of bronchial tree.

On the sample of 20 healthy volunteers times of propagation of the sounding signal to sensors located on the chest were studied. Evaluated times of arrivals were divided into groups by kernel density estimation plots. Obtained groups are represented in the form of "box and whiskers" diagram (Fig. 2).

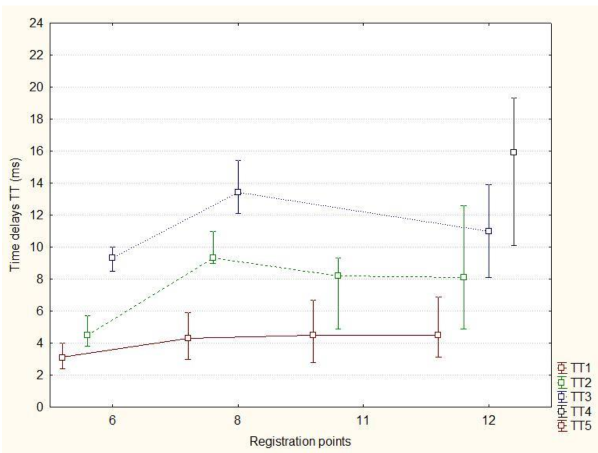

Fig. 2: Box and whiskers for obtained groups of propagation times under sounding through mouth, seen in >33% of the sample at left side of the chest surface

According to results of the Mann-Whitney test under sounding through mouth and supraclavicular chest areas there is a global symmetry of arrival times for sensors, located above the left and the right chest sides.

Using time delays and the distances between the sensors which were measured for all volunteers, we calculated a group speed of propagation. The velocities of sound waves are presented in diagram "box and whiskers" (Fig. 3) and they correspond to the propagation times (Fig. 2).

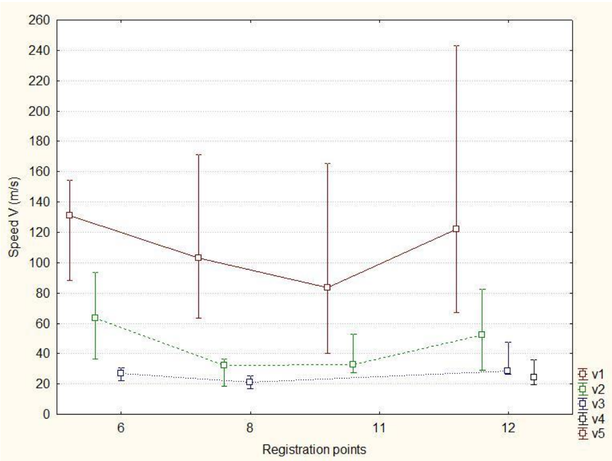

Fig. 3: Box and whiskers for obtained groups of propagation speeds under sounding through mouth, seen in >33% of the sample at left side of the chest surface

For physical reasons when sounding from the surface of the chest only structural mechanism of sound transmission is expected. In the structure of human chest, the sound may propagate by different ways, and its velocity depends on tissues characteristics and ways configuration.

The first arrivals with velocities of 70-160 m/s (right) and 100-200 m/s (left) most likely propagate through tissues with high density. While the rest of arrivals are apparently associated with a structural transmission through less dense tissues, in particular, through lung parenchyma, which as well known is characterized by a low sound speed of about 20 m/s.

On the contrary when a signal is applied through the mouth, both structural and air-structural mechanisms of propagation are possible [2]. The velocities of the 2nd arrivals under sounding through mouth are very close to the velocity of the 2nd and the 3rd arrivals under sounding from supraclavicular chest. Therefore these arrivals may be attributed to structural mechanism of transmission.

Whereas the velocities of the 1st arrivals under sounding through mouth are between 150-200 m/s (bronchial air lumen) and 20 m/s (lung parenchyma). Thus these arrivals may be attributed to air-structural mechanism of transmission.

Based on velocity estimates the 3rd arrival, when sounding through mouth, may be treated as a re-reflection of sound wave, propagating through air lumen, reflected from 11th-17th generations of bronchial tree, propagating in opposite direction, re-reflected from speaker diaphragm and then repeating the path of the 1st arrival from the speaker to the sensors on the chest surface by means of air-structural mechanism.

To assess the length of pass through lung parenchyma for air-structural transmission we can write the formula:

$$l_p(dT) = \{dT \cdot c_{air} - l_{air}\} \cdot \frac{c_p}{c_{air}}$$

where dT – propagation time, l_{air} – length of pass in air lumen, l_p – length of pass in lung parenchyma, c_p – speed of sound in parenchyma (20 m/s), c_{air} – speed of sound in air lumen (200 m/s).

An assessment of length of pass through lung parenchyma at various lengths of the way through the lumen of bronchial tree (from 4 to 24 cm with step 4 cm) showed that the structural propagation of the sound wave in lung parenchyma is only few centimeters whereas the lengths of pass through the airway lumen maybe almost an order longer. Consequentlythe air transmission mechanism is substantially more air-born, than thought previously [3].

III. Conclusions

When sounding through mouth and from supraclavicular chest areas with complex signal having bandwidth of 80-1000 Hz the results [1] about multiple sound transmissions to a chest wall and air-structural and structural mechanisms of propagation are confirmed in the extended sample and for different parts of the chest.

Global symmetry (left-right) of times of arrivals / group velocities is found under sounding through mouth as well as from supraclavicular chest areas.

It is shown that averaged group velocity of sound within the lumen of the bronchial tree does not exceed 150-200 m/s, and decreases from about 270 m/s in trachea as moving deeper in bronchial tree. It is confirmed independently that lengths of pass in air lumen are much more than in lung parenchyma and therefore air transmission mechanism is substantially more air-born, than was thought early [3].

Acknowledgments

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Determination of Lung Sound Spectral Parameters with and without Background Noise Subtraction

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Abstract—**The spectral edge frequency (SEF) is defined as the frequency below which contains 95% (SEF_{95}) or 99% (SEF_{99}) of the total spectral power between 150~1200Hz and we have shown that SEF of the inspiratory breath sound recorded at the chest surface is closely related with the degree of small airway obstruction measured with spirometry as $FEF_{25-75\%}$. When we determine the SEF in a flow targeted breath sound recording, we subtract an averaged breath-hold spectrum as background noises from an averaged inspiratory spectrum in order to get a net breath sound spectrum. Then, SEF is calculated with the net breath sound spectrum. However, it is often difficult to obtain a breath-hold spectrum in younger children. If the procedure of background noise subtraction (BNS) could be omitted without losing clinical value, application of lung sound analysis would become easier. Lung sounds were recorded in 30 known asthmatic children aged 6 to 15 years old, before and after salbutamol inhalation. $FEF_{25-75\%}$ were measured by spirometry just after each recording, and converted to the percent predicted values. We determined SEF_{95} and SEF_{99} with and without BNS and compared them in relation to $FEF_{25-75\%}$. Correlation with $FEF_{25-75\%}$ was highest in SEF_{99} with BNS (r=-0.658) and lowest in SEF_{99} without BNS (r=-0.409). But, in terms of correlation with $FEF_{25-75\%}$, SEF_{95} with and without BNS did not vary significantly (r=-0.647 vs r=-0.525). BNS is a desirable procedure in determination of lung sound spectral parameters. However, a parameter such as SEF_{95} is fairly insensitive to the omission of BNS procedure.**

Keywords— **children, bronchial asthma, vesicular sounds, spectral analysis, background noise**

I. INTRODUCTION

When we record breath sounds over the chest wall, contamination with non-respiratory sounds such as cardiovascular sounds and ambient noises is inevitable. These non-respiratory background noises could be determined by asking the subject to make a brief breath-hold at the end of recording. Thus, we can obtain a crude averaged breath sound spectrum at target flow rate and a non-respiratory background spectrum at zero flow rate. By subtracting the power of background spectrum from crude breath sound spectrum, a net breath sound spectrum could be determined. We have previously reported on several spectral parameters of inspiratory breath sound determined through application of this background noise subtraction (BNS) procedure [1-5]. The spectral edge frequency (SEF) defined as the frequency below which contains 95% (SEF_{95}) or 99% (SEF_{99}) of the total spectral power between 150~1200Hz, had been proved to be a sensitive indicator of airway narrowing in asthmatic children. SEF of the inspiratory breath sounds rises along with airway narrowing before it starts to wheeze. However, it is often difficult to obtain a breath-hold spectrum in younger children. If the procedure of BNS could be omitted without losing clinical value, application of lung sound analysis in younger children would become easier.

II. SUBJECTS AND METHOD

The subjects consisted of 30 known asthmatic children aged 6 to 15 years. They were tested under stable condition without any wheezes on auscultation. Lung sounds were recorded in a quiet room with a subject sitting upright wearing a nose clip and breathed through a pneumotach aiming to reach the target flow rate of 15ml/kg/s. They were instructed to make a ten second breath-hold at the end of recording. Two contact type sensors (Siemens, EMT25C) were placed at right upper chest and right lower back using double-sided adhesive ring tapes. The lung sounds were amplified and filtered through the custom made signal conditioner and digitized at a sampling rate of 10,240Hz through 12 bit AD converter equipped in a personal computer. Power spectra with 100ms segments were obtained through Fast Fourier Analysis (2048 point, with 50% overlap applying Hanning window). The average spectrum of the segments within a target flow range (+/-20%) was calculated and SEF without BNS was obtained. The net inspiratory spectrum was obtained by subtracting the average background noise spectrum, and SEF with BNS was obtained. As lung sounds were recorded simultaneously from two sites, the mean value of SEF indicated the individual value.

III. RESULTS

Correlation coefficients between $FEF_{25-75\%}$ and SEF_{99} or SEF_{95} with and without BNS were calculated (Table 1). With BNS, correlation between $FEF_{25-75\%}$ and SEF_{99} or SEF_{95} were comparable. However, without BNS, fall in the correlation was larger in SEF_{99} compared to SEF_{95}.

The relation between $FEF_{25-75\%}$ and SEF_{95} without BNS did not change significantly before and after bronchodilator (BD) inhalation (Fig.1). This is further confirmed by the strong correlation between the SEF changes with and without BNS following BD inhalation (Fig.2). As we have previously reported using SEF_{99} with BNS, the strong correlation between SEF at baseline and changes after BD inhalation was observed using SEF_{95} without BNS (Fig.3).

Table 1: Correlation coefficients between FEF25-75% and SEF with or without background noise subtraction (BNS)

Spectral index	BNS	Correlation coefficient
SEF_{99}	+	-0.658
	-	-0.409
SEF_{95}	+	-0.647
	-	-0.525

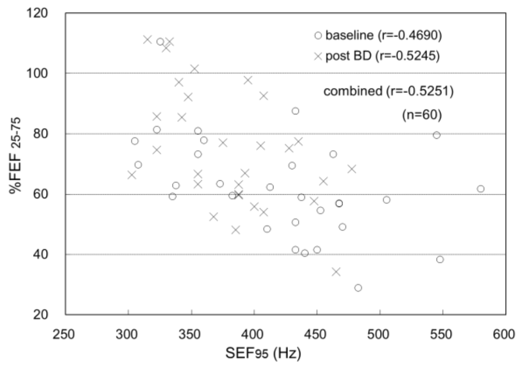

Fig. 1: Correlation between $\%FEF_{25\text{-}75}$ and SEF_{95} without BS before and after bronchodilator inhalation

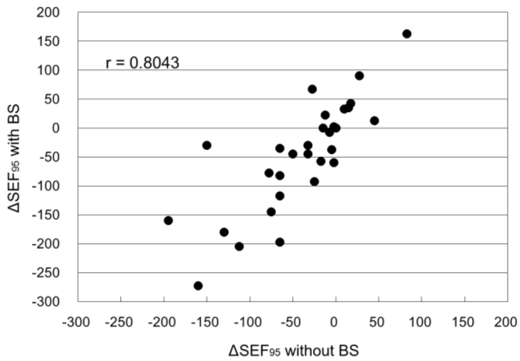

Fig. 2: Correlation between changes following bronchodilator inhalation in SEF95 with and without background subtraction

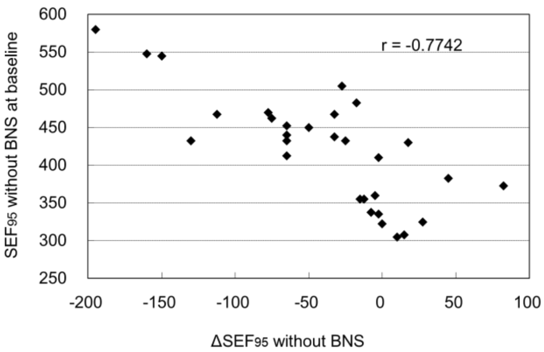

Fig. 3: Correlation between baseline SEF95 without BS and it's changes following bronchodilator inhalation

IV. Discussion

The result of this study indicate that the confounding effects of background noises may not be so destructive as far as a quiet recording environment is secured. However, as the present data were all recorded in the same environment, it does not ensure that the results here apply to the comparison of the spectral parameters recorded in the different environment.

There was a question about the percentage used in determining SEF, 99% or 95%. Our results showed that with the application of BNS, SEF_{99} may be slightly better than SEF_{95}, but SEF_{95} may be more robust to omission of BNS procedure.

In case of repeated measurement, such as those before and after bronchodilator inhalation, background noise may not change considerably. Although, change in heart rate could affect the value of SEF, heart sound effects are minimized by exclusion of spectral power below 150 Hz. Thus, it is not surprising that changes in SEF95 with and without BNS were highly correlated.

V. Conclusion

Distribution of a representative lung sound parameter SEF derived with and without BNS procedure were compared in relation with spirometric index $FEF_{25\text{-}75\%}$. Compared to SEF_{99}, SEF_{95} was less affected by omission of BNS procedure. SEF_{95} without BNS reflected changes following bronchodilator inhalation as good as the SEF_{95} with BNS. Thus, a parameter such as SEF_{95} is fairly insensitive to the omission of BNS procedure and likely to be used in younger children unable to hold their breath.

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Assessment Metrics & Performance Specifications for a Virtual Standardized Patient Comprehensive Pulmonary Auscultation Simulator

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Abstract— **Physical Examination skills in the United States are in decline relative to other countries. We have identified several error modes in the lung exam and establish a model to identify those errors within a high-fidelity virtual lung exam simulator and provide and remediation feedback as part of intentional practice in the pursuit of improved clinical pulmonary auscultation practices. This model is amenable to use in a virtual patient system. Improved auscultation skills have the potential to improve patient safety, lower costs and improve clinician satisfaction and confidence.**

Keywords— **Medical Simulation, Pulmonary Auscultation, Physical Examination, Medical Education.**

I. Introduction & methods

Physical examination (PE) skills among physicians in the United States are in an advanced state of decline compared to global standards [1]. Particularly impacted skills are pulmonary auscultation, the cardiac exam and the musculoskeletal exam. Reasons for this decline include lack of emphasis in training and dependence on technology-based testing [2]. As medical costs rise, there is an interest in restoring pulmonary auscultation skills [3]. The poor condition of these skills can be ameliorated with a targeted training program that includes proper examination technique modeling [4]. Regular refresher training is necessary to maintain these skills [5].

We previously described methods for simulating a high-fidelity virtual physical exam (VPE) whereby a multi-track audio program successfully demonstrated a dynamic sound mixing method with anatomical localization of sounds that was practical and subjectively mimicked reality [6]. The next phase in the design of a VPE is to identify error modes in the real-life performance of pulmonary auscultation so as to be able to detect such errors and provide corrective guidance in a future VPE simulator.

II. Results

A. Auscultation Exam Failure Modes

There are six errors common to physical exam procedures. The first basic error in pulmonary auscultation is failure to perform the procedure, something more common than physicians care to admit. The second error is the act of skipping parts of the exam which in the case of auscultation consists of listening at too few locations on the body. The third error is the use of a faulty examination technique. Examples of such auscultation errors include attempting to listen through clothing, failure to listen through a complete respiratory cycle, or failure to demand enough respiratory effort to provide adequate inspiratory and expiratory excursion. The fourth error is to fail to identify a physical finding. This error could be a failure to hear an abnormal sound, to recognize pathological sounds or identify the absence of sounds as a clinical finding. The fifth error is to misidentify a sound. Examples of this include confusing transmitted tracheal sounds as pulmonary rhonchi, identifying a crackle as a wheeze or a similar error. The sixth error is a failure to connect an auscultation finding with the correct diagnosis [3]. For this task, it is important to correlate auscultation discoveries with other physical exam findings and the patient interview record.

A landmark pulmonary auscultation skills study revealed that recent US medical graduates recognized only 40% of respiratory events in a standardized examination [7].

The root cause of these deficiencies is attributed to insufficient expert training and lack of exposure to abnormal findings [8]. Even when skills are developed, they fail to improve or decline in practice due to a lack of expert feedback and an increasing sense of uncertainty over time that leads physicians to depend more on radiological and laboratory tests.

Fig. 1: Virtual Standardized Patient (VSP) for Medical Interviewing

B. Assessment Design for a Lung Auscultation Simulator

We have already created a successful virtual standardized patient (VSP) simulator that is capable of conversational interviews in English, can provide actionable feedback for interview performance and can provide for a substantial training gain [9] [Figure 1]. Although we have been successful in methods to improve diagnostic interviewing skills, our physical exam presentation is very limited [Figure 2] and only provides an image or a description of a physical finding if it is selected from a list. This activity covers at most the first error mode whereby the learner must select the appropriate examination. In order to provide a complete learning experience, a simulation program must address as many error modes as possible within the confines of what can be performed on a modern computer or tablet. We can do better than the faux physical of the past, where a button was clicked to hear a recording. More importantly, expert feedback can provide for improved performance with repetitive intentional practice.

Fig. 2: VSP limited physical exam in current practice

A VPE system will have to expect the user to place a stethoscope on enough anatomical areas to evaluate all lobes and listen at locations where pathology is present. The VPE must detect if each auscultation site provided for at least one full respiratory loop and if excursion was adequate to detect findings; if not it should detect if the learner requested more excursion from the virtual patient. A VPE must allow for the learner to indicate that they discovered a finding or if the exam was perceived to be normal. If a finding were to be discovered, then the finding should be identified by the learner. Finally, the learner should link the finding to a clinical diagnosis.

All these VPE actions/errors are easy for a software simulation to detect and track as performance metrics. Of course, the VPE must provide high-fidelity lung sounds that vary by location of the stethoscope. Additionally, the sounds have to be temporally associated within the breathing loop, a loop which may vary based on learner requests for additional respiratory effort.

We have already prototyped the methods to provide for generalized, transmitted, localized and diminishing sounds [6] and have recently worked out a theoretical basis to tie these sounds to the respiratory loop. Such a potentially complex mixture of sounds, in our opinion, cannot easily be based on recordings, but will work better if synthetic sounds are employed to represent specific lung sounds, eliminating the problem of recording artifacts. We will model our sounds based on examples shared by members of the International Lung Sounds Association (ILSA).

Given the numerous actions being assessed, it is necessary to show a direct connection between the learner's actions and feedback provided by the VPE simulator, thus it will be necessary for a VPE to provide immediate live feedback at the time of an error or shortly thereafter. This provides for immediacy of feedback and allows the learner to associate their actions with a consequence in addition to discovering the correct action to take.

Upon misidentification, the feedback system must allow the learner to study the auscultation finding with its identity openly declared. Thus the learner can be reassured as to the identity of the sound that they are experiencing.

III. Conclusions

A virtual patient system can provide for standardized experiences and objective feedback [10]; two necessary components for "intentional practice", a well-understood method to achieve mastery in a domain. The combination of high-fidelity localized auscultation findings, an assessment system that can identify the six physical exam error modes, and a feedback capability that can assess, correct and remediate the learner has the potential to permit large gains in lung exam performance and instill clinicians with confidence in their skills so that they can save time, lower costs, and better serve patients. We will attempt to put these metrics and specifications into practice as we create our next VSP prototype. We will also attempt to study the efficacy of our approach. If we are successful, we hope clinicians will enjoy the pleasure of making a diagnosis with their senses a bit more often.

Acknowledgments

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Fine Crackles as Biomarker of Interstitial Pneumonia

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Abstract—**A male patient with interstitial pneumonia who experienced exacerbation of IP after cancer chemotherapy was studied. He was treated by steroid pulse and showed good response to this treatment. The sound power-spectrum of this case showed typical reversed S shape during exacerbation of interstitial pneumonia. After steroid pulse, this patient showed marked clinical improvement while the sound power-spectrum showed less prominent 1k Hz peak and reverse S shape. This sound-spectrographic change correlated well with his exertional dyspnea and exercise performance. Sound spectrographic analysis of fine crackles can be a useful biomarker in monitoring disease process of interstitial pneumonia.**

Keywords— **Interstitial pneumonia, crackles, sound spectrogram, lung sound, biomarker.**

I. Introduction

Fine crackles are brief, discontinuous high-pitched lung sounds. In ILSA2014, we reported a distinctive power spectrographic shape of fine crackles which has sound power peak around 1000 Hz and power dip at around 500Hz (reverse S shape). [1] We tried to clarify if this power spectrographic shape of fine crackles will be useful as a biomarker of interstitial pneumonia.

II. Subject and Method

A. Subject

A male patient in his sixties who had interstitial pneumonia (IP) and lung cancer was studied. His chest CT showed typical honeycombing. He experienced exacerbation of IP after cancer chemotherapy. We recorded and analyzed his lung sound since then. He was treated by steroid pulse therapy (Methyl prednisolone 1000mg/day x 3 days). He showed good response to this treatment.

B. Method

We analyzed breath sound by sound spectrometer (Kenz Medico, LSA 2010) and compared the frequency power-spectral shape of crackle sounds with chest X-ray and CT findings and clinical parameters, including six minutes walking test (6MWT). Size and condition of his lung cancer had been stable during this observation period.

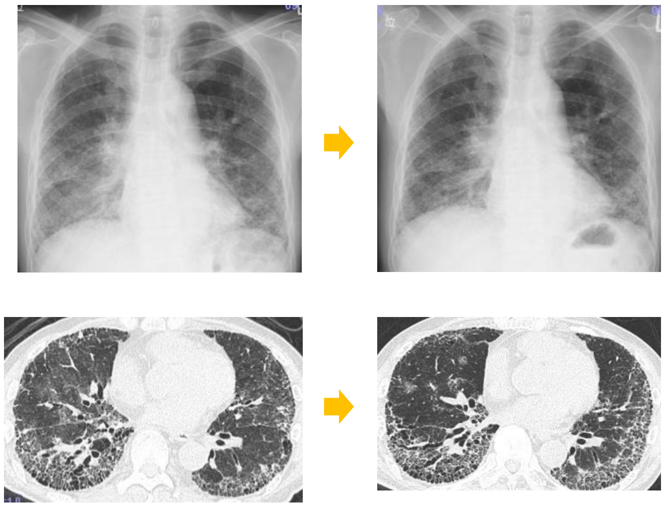

Fig. 1: Chest CT before and after steroid pulse for exacerbation of IP

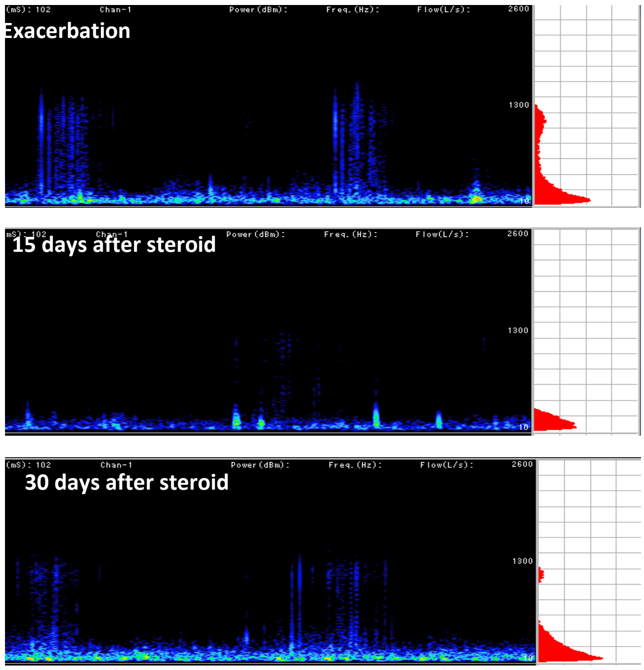

Fig. 2: Sound spectrogram of before and after steroid pulse

III. Results

Chest X-ray and CT showed reduction of diffuse pulmonary infiltrate after steroid pulse (Fig. 1). He noted improvement of exertional dyspnea after steroid pulse as shown in improvement of his exercise performance. Six minutes walking distance (6MWD) increased from 275m to 352m with less dyspnea. (Table 2) Blood examination showed marked decrease of KL6 and SP-D. (Table 1)

The frequency power-spectral shape of crackle sounds (Fig. 2) changed along with marked improvement in radiologic and clinical parameters.

Lung sound power spectrum during exacerbation showed typical reverse S shape, prominent round power peak around 1k Hz and dip around 500 Hz. The patient noted marked subjective improvement of exertional dyspnea 15 days after steroid pulse and at that time typical reverse S shape of power spectrum disappeared. Thirty days after steroid pulse, the patient noted a little increase of exertional dyspnea and there was some 1k Hz peak and 500 Hz dip in sound power spectrum. (Fig. 2)

Table 1: Blood parameters

	KL6	SP-D	LDH	CRP
Exacerbation	1250	233	365	0.38
15 days			342	0.49
30 days	342	0.49	342	0.49

Table 2: Lung functions and exercise parameters

	%FVC	%DLCO/Va	6MWD*	Borg scale**
Exacerbation	50	51	275 m	5/10
15 days	52	51	352 m	3/10
30 days	53	47	320 m	3/10

* 6MWD: six minutes walking distance
** Borg scale: Modified Borg scale.

IV. Discussions

Fine crackles are brief, discontinuous high-pitched lung sounds and have been defined by initial deflection width (IDW) and two cycle duration. [2] In ILSA2014, we reported a distinctive power spectrographic shape of fine crackles, reverse S shape. We showed that we will be able to distinct fine crackles from coarse crackles which show reverse J shape by sound spectrographic analysis. [1]

We tried to examine if this reverse S shape in sound power spectrum can be a marker of disease activity of IP. This case showed typical honeycombing in his chest CT and the diagnosis of interstitial pneumonia is definite. The disease process was closely monitored and an exacerbation and good response to steroid pulse therapy was observed. Although there was little improvement in lung function parameters after steroid pulse, good subjective improvement of exertional dyspnea and an increase in 6MWD was noted.

The height of sound power peak around 1k Hz or prominent reverse S shape correlated with the disease process of interstitial pneumonia. During our clinical observation, we noted that the change of lung sound was the most sensitive parameter of his degree of exertional dyspnea. As this in a single case, we are trying to analyze lung sounds in other cases of IP to see if this peculiar sound power spectrographic shape of fine crackles can be a good biomarker to monitor disease process and response to treatments in patients with IP.

V. Conclusions

The reversed S shape of sound power-spectrum correlated well with a disease process of interstitial pneumonia. We observed that the sound power spectrum correlated especially well with subjective symptoms such as exertional dyspnea and exercise performance. Power-spectrogram of fine crackles can be a useful biomarker in monitoring disease process of IP.

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