Human forced expiratory noise. Origin, apparatus and possible diagnostic applications\textsuperscript{a)

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ABSTRACT:
Forced expiratory (FE) noise is a powerful bioacoustic signal containing information on human lung biomechanics. FE noise is attributed to a broadband part and narrowband components—forced expiratory wheezes (FEWs). FE respiratory noise is composed by acoustic and hydrodynamic mechanisms. An origin of the most powerful mid-frequency FEWs (400–600 Hz) is associated with the 0th–3rd levels of bronchial tree in terms of Weibel ([2009]. Swiss Med. Wkly. 139(27–28), 375–386), whereas high-frequency FEWs (above 600 Hz) are attributed to the 2nd–6th levels of bronchial tree. The laboratory prototype of the apparatus is developed, which includes the electret microphone sensor with stethoscope head, a laptop with external sound card, and specially developed software. An analysis of signals by the new method, including FE time in the range from 200 to 2000 Hz and band-pass durations and energies in the 200-Hz bands evaluation, is applied instead of FEWs direct measures. It is demonstrated experimentally that developed FE acoustic parameters correspond to basic indices of lung function evaluated by spirometry and body plethysmography and may be even more sensitive to some respiratory deviations. According to preliminary experimental results, the developed technique may be considered as a promising instrument for acoustic monitoring human lung function in extreme conditions, including diving and space flights. The developed technique eliminates the contact of the sensor with the human oral cavity, which is characteristic for spirometry and body plethysmography. It reduces the risk of respiratory cross-contamination, especially during outpatient and field examinations, and may be especially relevant in the context of the COVID-19 pandemic. © 2020 Acoustical Society of America. https://doi.org/10.1121/10.0002705

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I. INTRODUCTION
Assessing human lung ventilation function is a standard approach for screening and diagnosing respiratory diseases, as well as monitoring effects of various extreme factors on the respiratory system, including diving, space flights, etc.

The common method of assessing human lung ventilation function is the flow-volume technique (spirometry and body plethysmography). Although filters and disposable or sterilized replaceable mouthpieces are used in the measuring equipment for testing pulmonary function, the risk of microbial contamination and cross-infection remains in this technique. It causes special restrictions on usage in current high-risk airborne respiratory infection during the COVID-19 pandemic (ERS, 2020). Moreover, the sensitivity of the traditional flow-volume technique of lung function diagnostics does not satisfy medical practice anymore. That is why attempts to develop alternative techniques for assessing lung ventilation function do not stop.

One of these attempts is connected to acoustic methods. Despite most of the studies in the field of respiratory acoustics being focused on the objectification of lung auscultation (see Rao et al., 2018; D’yachenko and Mikhaylovskaya, 2012), there are very few works aimed at developing acoustic methods for assessing lung ventilation function. A special breathing maneuver of forced exhalation, i.e., maximal sharp and complete exhalation after a full inspiration, similar to that performed during spirometry, is used frequently in these works.

Forced expiratory (FE) noise is a bioacoustic signal containing information on biomechanics of lungs (see, for example, Lyubimov et al., 2013). Furthermore, FE noise is characterized by high intensity and, consequently, a significantly higher signal-to-noise ratio than quiet breathing noise commonly used in medical auscultation of lungs. FE noise manifests via a powerful broadband part as well as by intensive narrowband components (see Korenbaum et al., 1998). The narrowband components were previously named as forced expiratory wheezes (FEWs) by Forgacs (1978).

Due to the mentioned reasons, FE noise is actively investigated for diagnostic purposes. An assumption about the possibilities of objectively analysing FEWs for diagnostics of bronchial obstruction was originally hypothesized by Forgacs (1978). Ishikawa et al. (1992) subsequently compared tracheal FEWs of 15 healthy individuals and 15 patients with asthma and found that sounds of healthy persons had an energy maximum in the frequency range from

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300 to 500 Hz. However, in asthma patients, a second spectral peak occurred in the frequency range between 1.3 and 1.5 kHz. Fiz et al. (2002) proposed the algorithm for automatic detection of FEWs in the three-dimensional spectrogram based on the results of the study by Shabtai-Musih et al. (1992). Fiz et al. (2006) demonstrated significant differences in the number of FEWs between asthma patients and healthy persons. However, diagnostic value of this parameter was insufficient. Korenbaum et al. (1998) also tried to use mid-frequency and high-frequency FEWs for diagnostics of asthma and chronic obstructive pulmonary disease (COPD); however, isolated evaluation of FEWs was found to be diagnostically ineffective.

Several authors have analysed FEWs in healthy individuals. Charbonneau et al. (1987) found that FEWs in healthy persons appeared after achievement of maximum flow velocity, but sometimes with essential delay. FEWs were also associated with certain forms of the flow-volume curve. Beck and Gavriely (1990) showed that FEWs registered above the trachea in healthy individuals were produced in 95% of all maneuvers. Moreover, in various maneuvers of each person through a tube of the same diameter (i.e., in identical flow conditions), the main component of the FEWs was characterized by a similar peak frequency in all attempts.

Nevertheless, the mechanisms of FE noise production and the levels of bronchial tree responsible for these noise origins, as well as possibilities to use parameters of FE noise for diagnosis and monitoring of human lung function, are still being discussed.

The objective of this work is to clarify acoustic considerations about the origin and characteristics of FE noises registered over trachea, to develop an acoustic technique for registering and processing FE noises, and to draft possible applications of this technique for diagnostics and monitoring of human lung function.

II. ACOUSTIC CONSIDERATIONS ON ORIGIN OF FE NOISES

A simplified scheme demonstrating FE signal origin and its recording above human trachea is shown in Fig. 1. During sound signal registration by a microphone, equipped with a stethoscope head, FE tracheal noises may appear in two ways (see D’yachenko et al., 2012; Glazova et al., 2018). The first one is a superposition of acoustic noises emitted into a lumen of airways. This mechanism involves a propagation of acoustic signals from distant intrabronchial sources (Fig. 1) through airways of the bronchial tree, resulting in formation of an acoustic pressure ($p_a$) inside the trachea lumen. The second mechanism is hydrodynamic or so-called pseudo-sound effect of turbulent pressure pulsations of airflow vortices to the tracheal inner wall. This effect results in an averaged hydrodynamic pressure ($p_{hd}$). Unlike the first (acoustic) mechanism, there is no need for air medium compressibility here, and the recorded signal is proportional to the hydrodynamic pressure in the turbulent flow averaged through an area of perception of the acoustic sensor. Due to the physics of its operation, the stethoscope sensor on the neck does not distinguish these mechanisms of pressure changes on the tracheal inner wall and registers equally both signal components outside trachea (see Korenbaum et al., 2016b).

As one can see in the spectrogram (Fig. 2), FE noises are represented by the powerful broadband part and intensive narrowband FEWs. According to the spectrogram (Fig. 2), off-prints of some types of FEWs may be identified. The first one—a path of the most prominent mid-frequency FEW seen usually during almost all time of FE maneuver in the frequency range from 400 to 600 Hz (1), the second one—a path of the early high-frequency FEW seen in the first half of FE maneuver in the frequency range over 600 Hz (2), the third one—a path of the late high-frequency FEW seen in the second half of FE maneuver in the frequency range over 600 Hz. Although peak frequencies of FEWs vary essentially between subjects, the scheme of FEWs appearance, described in Fig. 2 may be found above trachea in the majority of volunteers and patients (Korenbaum et al., 2013b).

Korenbaum et al. (1997) and Korenbaum et al. (2009) supposed that FEWs may be divided into sounds generated by flow-dependent mechanisms (vortices shedding, forced
TABLE I. The list of developed band-pass energies and durations in the 200 Hz bands, where $A = \Sigma A_i$.

| $i$ | Frequency band, Hz | $A_i$, conventional unit | $t_{ri}$, second | $A_{ri}$, fraction of $1$ | $t_{ri}$, fraction of $1$ |
|-----|-------------------|--------------------------|-----------------|-------------------------|--------------------------|
| 1   | 200–400           | $A_{200–400}$            | $t_{200–400}$   | $A_{200–400}/A$         | $t_{200–400}/FET_a$      |
| 2   | 400–600           | $A_{400–600}$            | $t_{400–600}$   | $A_{400–600}/A$         | $t_{400–600}/FET_a$      |
| 3   | 600–800           | $A_{600–800}$            | $t_{600–800}$   | $A_{600–800}/A$         | $t_{600–800}/FET_a$      |
| 4   | 800–1000          | $A_{800–1000}$           | $t_{800–1000}$  | $A_{800–1000}/A$        | $t_{800–1000}/FET_a$     |
| 5   | 1000–1200         | $A_{1000–1200}$          | $t_{1000–1200}$ | $A_{1000–1200}/A$       | $t_{1000–1200}/FET_a$    |
| 6   | 1200–1400         | $A_{1200–1400}$          | $t_{1200–1400}$ | $A_{1200–1400}/A$       | $t_{1200–1400}/FET_a$    |
| 7   | 1400–1600         | $A_{1400–1600}$          | $t_{1400–1600}$ | $A_{1400–1600}/A$       | $t_{1400–1600}/FET_a$    |
| 8   | 1600–1800         | $A_{1600–1800}$          | $t_{1600–1800}$ | $A_{1600–1800}/A$       | $t_{1600–1800}/FET_a$    |
| 9   | 1800–2000         | $A_{1800–2000}$          | $t_{1800–2000}$ | $A_{1800–2000}/A$       | $t_{1800–2000}/FET_a$    |

Thus a variety of identified acoustic FE effects opens new opportunities for diagnostic applications.

III. METHOD AND APPARATUS

The acoustic method of diagnostics of lung function is developed. It is based on the estimation of noise parameters of human forced exhalation, recorded above trachea (Fig. 1). As assessing FEWs parameters turned out to be technically difficult, alternative band-pass acoustic parameters of the FE noise were developed. Taking into account a dependence of FE noise amplitude characteristics on properties of acoustic tracts, special attention was paid to develop parameters independent of the sensitivity of sensors and amplification of electronics.

The first parameter suggested was FE noise time in the frequency range of 200–2000 Hz (FET$_a$), which characterized the noise process in common. A frequency band below 200 Hz was excluded due to high level of interference associated with the vibration of an acoustic sensor when touching soft tissues. A frequency band above 2000 Hz was excluded due to the low level of the signal. The FET$_a$ is less dependent on properties of acoustic tracts than any amplitude characteristics.

Measurement of FET$_a$ for each recorded file was taken by using a specially developed algorithm. Filtration was carried out in the frequency band of 200–2000 Hz (Kaiser windowed direct-form finite impulse response, FIR, filter). The FE waveform envelope was constructed doubly in the forward and opposite directions by moving the average method with an accumulation period of 0.01 s. Then the peak amplitude ($U$) of the envelope was calculated. The threshold level $S = 0.005U$ was defined. The times of beginning $T_1$ and ending $T_2$ of the FE noise process were measured by the threshold level $S$ of the envelope when moving from the peak to the left and to the right. Since $T_1$, $T_2$ were measured automatically or semiautomatically (see Korenbaum et al., 2013a), the program automatically calculated the FE noise time as the difference $FET_a = T_2 - T_1$. The maximal individual FET$_a$ from at least three well-done attempts of forced exhalations was used for further analysis.

Additionally, the band-pass durations and energies in the bands of 200 Hz lying inside the total frequency range 200–2000 Hz are developed supposedly more suitable for estimating narrow-band FE noise parameters than FEWs. They are specified in Table I.

The special algorithm is designed to calculate $A_i$ and $t_i$ shown as an example in Fig. 3.

The band-pass durations are measured using the envelope of the signal filtered in each 200 Hz frequency band, constructed by moving average method, by means of summing the number of all time samples of the signal in which amplitudes exceeded the threshold $S$. To evaluate the band-pass durations, the number of fixed time samples is multiplied by the period between two samples of the envelope (0.01 s). The band-pass energies are calculated using the same envelope. Here, the amplitudes of fixed time samples of the signal are summed if their magnitudes exceed the threshold $S$. The sections of the envelope of the signal lying below the threshold $S$ are excluded in both cases. Vice versa, all fragments of the envelope of the signal lying above the threshold $S$ are summed as $A_i = A_k + A_{m}$, and $t_i = t_k + t_m$ (Fig. 3). The threshold $S = 0.005U$ is the same as used in FET$_a$ measuring procedure.

Finally, normalized band-pass durations $t_{ri} = t_i/FET_a$ and normalized band-pass energies $A_{ri} = A_i/A$ $(i = 1, \ldots , 9)$ are calculated, where $A = \Sigma A_i$ is total energy in the frequency range of 200–2000 Hz. Note that normalized band-pass parameters, in contrast to absolute ones, are relative values, thus insensitive to variations in the transfer coefficient of sensors and amplifiers. It makes these acoustic parameters more convenient for diagnostic usage.

FIG. 3. Example of evaluating the band-pass duration $t_i$ and the band-pass energy $A_i$ by the envelope of FE noise signal filtered in the $i$th 200 Hz band.
To illustrate typical values for the developed parameters in all 200-Hz bands as well as their repeatability in the form of the coefficient of variation, $CV = SD/M$. Table II is included, summarizing data for one healthy male subject collected in his four attempts of FE maneuver.

The laboratory prototype of apparatus is developed, which includes an acoustic sensor–electret microphone (W62A) with an ebonite stethoscope head. The head has a conical chamber of 20 mm diameter at its base and 5 mm in depth (opening angle of 120°). A laptop provided with an external sound card Transit (M-Audio) and specially developed software is used to introduce signal from the microphone and to evaluate $FET_a$ (see Korenbaum et al., 2013a) and the 200-Hz band-pass durations and energies in an automatic manner.

During measurements, the sensor is attached to the lateral neck surface and the subject or physician (operator) holds the box with his hand pressing a stethoscope from head to the body. Nose-clamp is used. The subject performs a forced exhalation maneuver from maximal inspiration. A delay of about 0.5 s is made between inspiration and exhalation. In order to carry out the maneuver properly, a maximum sharp and complete exhalation is required (see Pocheutkova and Korenbaum, 2013).

To assess possible diagnostic applications of the developed acoustic method, let us analyse now the results of processing FE noise signals previously collected by our team in volunteers and patients.

IV. RESULTS AND DISCUSSION OF SOME DIAGNOSTIC APPLICATIONS

The first question is whether the developed acoustic parameters reflect traditional biomechanical flow-volume indices of human pulmonary function.

The main method of study of acoustical-biomechanical relations was a comparison of the FE tracheal noise acoustic parameters with the results of the evaluation of biomechanical indicators of human lung function, which were obtained not only by means of spirometry but also with body plethysmography (MasterScreen Body, Jager).

Korenbaum et al. (2016a) studied the sample consisting of 218 volunteers, which included healthy subjects ($n = 50$), persons with risk factors of chronic respiratory diseases ($n = 60$), patients with spirometry negative ($n = 32$), and spirometry positive ($n = 41$) bronchial asthma and COPD ($n = 35$). It is evident from clinical considerations that the factor of “incidence and severity” of bronchial obstruction increases gradually in the sample from the 1st group to the 5th one. By means of nonparametric Jonckheere-Terpstra analysis of variance (ANOVA) analysis, a statistically significant ($p < 0.001$) effect of the factor of “incidence and severity” of bronchial obstruction on lung function biomechanical indices is revealed, for example, on the body plethysmographic airway resistance [Fig. 4(a)]. Thus, now using the sample characterized by the significant gradual increase in the body plethysmographic airway resistance [Fig. 4(a)] as the model, one could see that $FET_a$ [Fig. 4(b)] is well coordinated with the airway resistance (Jonckheere-Terpstra test, $p < 0.001$). This observation confirms previously developed model considerations of FE noise production in healthy individuals, in individuals with bronchial obstruction (see Korenbaum and Pocheutkova, 2008), and in volunteers breathing gases of various density (see D’yachenko et al., 2012), connecting $FET_a$ with indirect measures of FE airway resistance.

Furthermore, Pocheutkova and Korenbaum (2013) demonstrated for the $FET_a$ acoustic parameter a sufficiently effective acoustic diagnosis of bronchial obstruction in patients with spirometry confirmed asthma (group 4 in Fig. 4) with sensitivity and specificity of about 90%, as well as a possibility of detection of hidden bronchial obstruction not revealed by traditional spirometry in near 50% of the group 3 (Fig. 4) with spirometry negative asthma.

Malaeva et al. (2017) revealed significant interrelations between band-pass acoustic durations and energies of the FE tracheal noises and biomechanical indices of lung

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**TABLE II.** An example of a data set of the healthy male subject averaged in four attempts of FE maneuver.

| Parameter | Units | $M$ | CV, % |
|-----------|-------|-----|-------|
| $FET_a$   | s     | 1.441 | 1.7 |
| $t_{200-400}$ | s     | 1.121 | 4.5 |
| $t_{400-600}$ | s     | 1.262 | 2.5 |
| $t_{600-800}$ | s     | 1.180 | 9.1 |
| $t_{800-1000}$ | s   | 1.246 | 6.7 |
| $t_{1000-1200}$ | s  | 1.325 | 5.0 |
| $t_{1200-1400}$ | s | 1.185 | 1.6 |
| $t_{1400-1600}$ | s  | 0.807 | 17.2 |
| $t_{1600-1800}$ | s  | 0.691 | 31.0 |
| $t_{1800-2000}$ | s  | 0.575 | 11.0 |
| $A_{200-400}$ | Conventional | 19.082 | 43.6 |
| $A_{400-600}$ | Conventional | 19.711 | 32.7 |
| $A_{600-800}$ | Conventional | 14.092 | 40.9 |
| $A_{800-1000}$ | Conventional | 25.198 | 39.5 |
| $A_{1000-1200}$ | Conventional | 31.651 | 30.3 |
| $A_{1200-1400}$ | Conventional | 17.786 | 24.4 |
| $A_{1400-1600}$ | Conventional | 5.362 | 38.6 |
| $A_{1600-1800}$ | Conventional | 3.196 | 29.0 |
| $A_{1800-2000}$ | Conventional | 1.714 | 25.6 |
| $t_{200-400}$ | Fractions of 1 | 0.778 | 4.5 |
| $t_{400-600}$ | Fractions of 1 | 0.876 | 2.5 |
| $t_{600-800}$ | Fractions of 1 | 0.819 | 8.6 |
| $t_{800-1000}$ | Fractions of 1 | 0.865 | 7.4 |
| $t_{1000-1200}$ | Fractions of 1 | 0.919 | 5.0 |
| $t_{1200-1400}$ | Fractions of 1 | 0.823 | 2.6 |
| $t_{1400-1600}$ | Fractions of 1 | 0.559 | 16.0 |
| $t_{1600-1800}$ | Fractions of 1 | 0.478 | 30.3 |
| $t_{1800-2000}$ | Fractions of 1 | 0.399 | 10.2 |
| $A_{200-400}$ | Fractions of 1 | 0.137 | 15.6 |
| $A_{400-600}$ | Fractions of 1 | 0.148 | 11.0 |
| $A_{600-800}$ | Fractions of 1 | 0.103 | 13.3 |
| $A_{800-1000}$ | Fractions of 1 | 0.183 | 15.3 |
| $A_{1000-1200}$ | Fractions of 1 | 0.241 | 17.9 |
| $A_{1200-1400}$ | Fractions of 1 | 0.136 | 12.2 |
| $A_{1400-1600}$ | Fractions of 1 | 0.040 | 13.7 |
| $A_{1600-1800}$ | Fractions of 1 | 0.024 | 16.5 |
| $A_{1800-2000}$ | Fractions of 1 | 0.013 | 20.1 |
function by means of nonparametric Spearman correlation analysis in the sample. The strongest correlation coefficients are noted between band-pass durations and measures of airway resistances (up to 0.59), reflecting primarily the function of large airways, as well as the residual volume of the lungs (up to 0.47) and its ratio to the total lung capacity (up to 0.43) that characterize the state of small airways. The significant bidirectional correlation between acoustic FE tracheal parameters and biomechanical indices were additionally revealed in each specific group of the sample.

The analyzed findings mean that developed FE acoustic parameters correspond to basic indices of human lung function measured by spirometry and body plethysmography, and in some cases (for example, spirometry negative asthma), may be even more sensitive to respiratory deviations than mentioned indices.

To assess some possibilities of the acoustic estimation of the impact of various extreme factors to the human respiratory system, we studied diving submersions and postural modeling of microgravity.

Pochekutova and Korenbaum (2011) applied the developed acoustic technique to divers (48 subjects) and revealed an increase of FET_a interpreted as the acoustic sign of transient bronchial obstruction features in 13 subjects (27%) after single shallow-water sea submersion with the old fashioned closed-circuit breathing apparatus IDA-71 (SU). The effect was probably caused by the development of inflammation of bronchial mucosa and accompanying edema due to the toxic effect of hyperbaric hyperoxia in combination with small doses of the regenerative substance (including caustic potassium alkali) vapor. These signs of toxic damage to the pulmonary system appeared in time intervals not exceeding the permissible period of the diving operation with oxygen. The observation dictated a necessity to provide individual control of divers’ lung function during the training process in a closed-circuit breathing apparatus in order to prevent accidents and to achieve professional longevity.

It is supposed that a modern closed-circuit breathing apparatus would not have such influence on human lung function due to an absence of the regenerative substance, including caustic potassium alkali. To test this supposition, a group of 25 male divers performed single shallow-water sea submersion (less than 1 h) in a modern closed-circuit breathing apparatus FROGS (AquaLung, France) (see Table III). Although there is no significant rise of FET_a in relation to background status, a significant decrease of normalized band-pass high-frequency energy \( \text{Ar}_{1400-1600} \) is revealed (Wilcoxon \( p = 0.03 \)). It is interesting that in the referent group of healthy volunteers (\( n = 29 \)), consisting of 16 males and 13 females under a medical salbutamol bronchodilator test (Table III), the opposite significant response of the adjacent normalized band-pass high-frequency energy \( \text{Ar}_{1600-1800} \) is observed (Wilcoxon \( p = 0.017 \)). Thus the response found in divers may be treated as an adverse influence of even a short oxygen submersion on increasing airway resistance. This effect is consistent with those obtained for the closed-circuit breathing apparatus of the previous generation (Pochekutova and Korenbaum, 2011). It looks like the phenomenon revealed does not depend on the configuration

![Diagram](image.jpg)

**FIG. 4.** Diagrams of the dependence of body plethysmographic airway resistance \( R_{ex} \) (a), and FET_a (b) on the “incidence and severity” of bronchial obstruction in specific groups of the sample of 218 persons (Jonckheere-Terpstra test, \( p < 0.001 \) both for \( R_{ex} \) and FET_a); 1, healthy volunteers, 2, subjects with risk factors of bronchial obstruction; 3, spirometry negative asthma patients; 4, spirometry positive asthma patients; 5, COPD patients.

| Studied parameter | Underwater submersion (\( n = 25 \)) | Medical bronchodilator test (\( n = 29 \)) |
|------------------|-------------------------------------|------------------------------------------|
| FET_a            | ns                                  | ns                                       |
| \( \text{Ar}_{1400-1600} \) | 0.03 (−)                           | ns                                       |
| \( \text{Ar}_{1600-1800} \) | ns                                  | 0.017 (+)                                |
| \( \text{Ar}_{1800-2000} \) | ns                                  | ns                                       |
of the closed-circuit breathing apparatus, but is rather defined by hyperbaric hyperoxia itself. This conclusion seems to coordinate well with the results of a recent study by Wingelaar et al. (2019), which involved comparative analysis of volatile organic compounds in exhaled air after one-hour underwater dive using air and oxygen. In oxygen divers, a significant increase in the content of volatile compounds (methyl alkanes) is revealed, which is connected with damage of phosphatidylcholine membranes of pulmonary structures by hyperbaric hyperoxia. Nevertheless, flow-volume indices of spirometry have not changed yet. Therefore, our current study found that acoustic signs of an increase in airway resistance may be treated as a result of similar damage/inflammation reaction in the human respiratory tract.

As for microgravity simulation, Malaeva et al. (2018) previously studied postural simulation of microgravity by means of head-down −6° test (exp_1, five males) and lunar gravity by means of head-up +9.6° test (exp_2, six males) during a 20-day experiment. The reason for postural changes was a simulation of a lunar mission with part of a crew in an orbital device (microgravity) and another part on the Moon’s surface (lunar gravity). Both subgroups experienced five days in the head-down position (simulating space flight). On the fifth day, the second subgroup was transferred into the head-up position (simulating landing on the Moon). It was revealed that FETa could distinguish subgroups subjected to these impacts from the 6th–20th days of the experiment (2-factor ANOVA), while there was no essential response of basic spirometry indices found. Furthermore, FETa in prolonged simulation of microgravity was significantly higher than in prolonged simulation of lunar gravity. Using acoustical-biomechanical model considerations, an increase of FETa in the microgravity model may be explained by an additional rise in airway resistance with respect to the lunar gravity model.

The developed normalized band-pass energies are applied additionally to analyze the results of that experiment. It is found that relative band-pass energies \( A_{r800-1000} \) and \( A_{r1400-1600} \) (Fig. 5) could distinguish studied subgroups (exp_1 vs exp_2) in the time interval from the 6th to 20th days of the experiment (2-factor ANOVA) no less successfully than FETa in Malaeva et al. (2018).

It is interesting that directions of responses of these newly developed parameters to long-term postural effects are opposite (Fig. 5). Moreover, in accordance with the results of the Least Significant Difference (LSD)-test (Table IV), the first parameter significantly responds to postural change immediately on the sixth day of the experiment, while the second one reacts later—only by the ninth day of the experiment.

These findings open the possibility to more detailed acoustic-biomechanical interpretation in studied postural models. On the other hand, the results seem promising for the implementation of subtle individual control of human pulmonary function in microgravity conditions.

Consequently, based on preliminary results of analyzed diagnostic applications, the developed technique may be considered a promising instrument for solving the problem of acoustic monitoring of human lung function status in various extreme conditions, including diving and space flights.

Additional studies connected to more extensive exploration of the developed set of FE acoustic parameters (Table I) in clinical testing of wider samples of volunteers and respiratory patients are welcome.

The developed technique has a sufficiently high sensitivity to detect deviations of human lung ventilation function. Its findings, as a rule, coordinate well with flow-volume TABLE IV. Statistical significance (\( p \)) of distinctions of the normalized band-pass energies (Fig. 5) between the model of microgravity (exp_1) and model of microgravity/lunar gravity (exp_2) in the same days of the experiment according to LSD test. ns, \( p < 0.05 \).

|               | 3rd | 6th | 9th | 14th | 20th | After |
|---------------|-----|-----|-----|------|------|-------|
| \( A_{r800-1000} \) | ns  | ns  | 0.004 | 0.0007 | 0.02 | 0.0009 | ns    |
| \( A_{r1400-1600} \) | ns  | ns  | ns   | 0.009 | 0.002 | 0.02 | ns    |
pulmonary indices. However, the developed technique eliminates the contact of the sensor with the oral cavity, which is characteristic of spirometry and body plethysmography. It reduces the risk of respiratory cross-contamination, especially during outpatient and field examinations, and may be especially relevant in the context of the COVID-19 pandemic.

V. CONCLUSIONS

An origin of FE noise is detailed in relation to mechanisms of the broadband part as well as narrowband FEWs. Informative acoustic parameters of FE noise, suitable for diagnostics of lung function, are found. The laboratory prototype of the apparatus is developed, which includes an electret microphone with stethoscope head, a laptop with an external sound card, and specially developed software. The FE time in the range of 200–2000 Hz and the band-pass durations and energies in 200 Hz bands are evaluated instead of FEW direct measures. It is demonstrated experimentally that developed FE acoustic parameters correspond to basic indices of lung function evaluated by spirometry and body plethysmography and may be even more sensitive to respiratory deviations. According to preliminary experimental results, the developed technique may be considered a promising instrument for acoustic monitoring of human lung function in extreme conditions, including diving and space flights. It is important that, in comparison with traditional flow-volume methods, the developed technique reduces the risk of respiratory cross-contamination.

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Beck, R., and Gavriely, N. (1990). “The reproducibility of forced expiratory wheezes,” Am. Rev. Respir. Dis. J. 141, 1418–1422.

Charbonneau, G., Sudraud, M., Racineux, J. L., Meslier, N., and Tuchais, E. (1987). “Is the shape of the flow rate curve related to existence of a wheeze?,” Chest 92, 825–831.

D’yachenko, A. I., Korenbaum, V. I., Shulagin, Y. A., Osipova, A. A., Mikhailovskaya, A. N., Popova, Y. A., Kryanova, E. V., and Pochekutova, I. A. (2012). “Influence of modified gas mixtures on the acoustic parameters of human forced exhalation,” Human Physiol. 38(1), 77–83.

D’yachenko, A. I., and Mikhailovskaya, A. N. (2012). “Respiratornyaya akustika (obzor)” (“Respiratory acoustics (review),” Trudy IFOFAN 68, 136–181 (in Russian).

ERS (2020). “Lung function testing during COVID-19 pandemic and beyond—Group. 9.1. Respiratory function technologists/scientists,” https://www.ersnet.org/covid-19-guidelines-and-recommendations-directory (Last viewed May 29, 2020).

Fiz, J. A., Jene, R., Homls, A., Izquierdo, J., Garcia, M. A., and Morera, J. (2002). “Detection of wheezing during maximal forced exhalation in patients with obstructed airways,” Chest 122, 186–191.

Fiz, J. A., Jene, R., Izquierdo, J., Homls, A., Garcia, M. A., Gomez, R., Mones, E., and Morera, J. (2006). “Analysis of forced wheezes in asthma patients,” Respiration 73, 55–60.

Fogacs, P. (1978). “The functional basis of pulmonary sounds,” Chest 73, 399–405.

Glazova, A. Y., Korenbaum, V. I., Kostiv, A. E., Kabancova, O. I., Tagiltcev, A. A., and Shin, S. N. (2018). “Measurement and estimation of human forced expiratory noise parameters using a microphone with a stethoscope head and a lapel microphone,” Physiol. Meas. 39(6), 065006.

Ishikawa, S., Beauchamp, H., Kenney, L., Arason, R., Allard, J., McNalty, S., Krashim, M., and Macdonnell, K. F. (1992). Proceedings of the 17th International Conference on Lung Sounds, August 24–26, Helsinki, Finland.

Korenbaum, V. I., Kulakov, J. U., and Tagiltcev, A. A. (1997). “Acoustic effects in the human respiratory system under forced expiration,” Acoust. Phys. 43(1), 66–73.

Korenbaum, V. I., and Pochekutova, I. A. (2008). “Regression simulation of the dependence of forced expiratory tracheal noises duration on human respiratory system biomechanical parameters,” J. Biomech. 41(1), 63–68.

Korenbaum, V. I., Pochekutova, I. A., Kostiv, A. E., Tagiltcev, A. A., and Shubin, S. B. (2013a). “Technology of human pulmonary function testing by means of tracheal forced expiratory noises analysis,” IFMBE Proc. 39, 2192–2195.

Korenbaum, V. I., Pochekutova, I. A., Malaya, V. V., and Kostiv, A. E. (2016a). “Acoustic biomechanical relationships of human forced exhalation in bronchial obstruction,” Hum. Physiol. 42(4), 421–424.

Korenbaum, V. I., Rasskazova, M. A., Pochekutova, I. A., and Fershalov, Y. V. (2009). “Mechanisms of sibilant noise formation observed during forced exhalation of a healthy person,” Acoust. Phys. 55(4–5), 528–537.

Korenbaum, V. I., Safronova, M. A., Markina, V. V., Pochekutova, I. A., and D’yachenko, A. I. (2013b). “Study of the formation mechanisms of forced expiratory wheezes in a healthy person when breathing gas mixtures of different density,” Acoust. Phys. 59(2), 240–249.

Korenbaum, V. I., Tagiltcev, A. A., Goroyov, S. V., Shiryaev, A. D., and Kostiv, A. E. (2016b). “On localization of wheezing respiratory sounds in human lungs by means of intemismetric processing of signals detected on the chest surface,” Acoust. Phys. 62(5), 600–607.

Korenbaum, V. I., Tagiltcev, A. A., Kulakov, J. U., Kilin, A. S., Avdeeva, H. V., and Pochekutova, I. A. (1998). “An acoustic model of noise production in the human bronchial tree under forced expiration,” J. Sound Vib. 213(2), 377–382.

Lyubimov, G. A., Skobeleva, I. M., Dyachenko, A. I., and Strongin, M. M. (2013). “Estimation of the forced expiration tracheal sounds intensity,” Physiol. Cheloveka 39, 126–134 (in Russian).

Malaya, V. V., Korenbaum, V. I., Pochekutova, I. A., Kostiv, A. E., Shin, S. N., Katuntsev, V. P., and Baranov, V. M. (2018). “A technique of forced expiratory noise time evaluation provides distinguishing human pulmonary ventilation dynamics during long-term head-down and head-up tilt bed rest tests simulating micro and lunar gravity,” Front. Physiol. 9, 1255.

Malaya, V. V., Pochekutova, I. A., Kostiv, A. E., Shin, S. N., and Korenbaum, V. I. (2017). “Correlation between acoustic characteristics of forced expiratory tracheal noises and lung function parameters in healthy subjects and patients with obstructive lung diseases,” Hum. Physiol. 43(6), 662–669.

Pochekutova, I. A., and Korenbaum, V. I. (2011). “Acoustic estimation of the impact of a single dive using a closed-type breathing apparatus on the ventilatory function of the human lungs,” Hum. Physiol. 37, 334–338.

Pochekutova, I. A., and Korenbaum, V. I. (2013). “Hidden bronchial obstruction diagnostics by means of computer assessed tracheal forced expiratory noise time,” Respiratory 18(3), 501–506.

Rao, A., Ruiz, J., Bao, C., and Roy, S. (2018). “Acoustic methods for pulmonary diagnosis,” IEEE Rev. Biomed. Eng. 12, 221–239.

Shabtai-Musiy, Y., Grotberg, J. B., and Gavriely, N. (1992). “Spectral content of forced expiratory wheezes during air, He, and SF6 breathing in normal humans,” J. Appl. Physiol. 72, 629–635.

Weibel, E. R. (2009). “What makes a good lung? The morphometry basis of lung function,” Swiss Med. Wkly. 139(27–28), 375–386.

Wetlauer, T. T., van Ooij, P.-J. A. M., Brinkman, P. J. A. M., and van Hutst, R. A. (2019). “Pulmonary oxygen toxicity in navy divers: A crossover study using exhaled breath analysis after a one-hour air or oxygen dive at nine meters of sea water,” Front. Physiol. 10, 10.