Aeroacoustic sound alteration in airway bronchoconstriction, represented by a constricted T-branch model

Gabriel Pramudita SAPUTRA*, Kazunori NOZAKI**, Satoshi II* and Shigeo WADA*

*Graduate School of Engineering Science, Osaka University
1-3 Machikaneyama, Toyonaka, Osaka 560-8531, Japan
E-mail: shigeo@me.es.osaka-u.ac.jp
**Division of Dental Informatics, Osaka University Dental Hospital
1-8 Yamadaoka, Suita, Osaka 565-0871, Japan

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Abstract

Lung sound is commonly analyzed when testing for lung abnormalities. However, the accuracy of this analysis is low and more information on sound generation and alteration is needed to improve the analysis quality. In the current study, an aeroacoustic investigation of sound generation in an airway model was performed experimentally to uncover the factors that change the sound characteristics due to bronchoconstriction. A T-branch configuration was used to represent an airway junction, and a constricted tube in the mother branch was used to represent a bronchoconstriction. Aeroacoustic sound was analyzed at several flow rates, representing different speed maneuvers. Constriction percentages of 25%, 50%, and 75% simulated different bronchoconstriction severities. The power spectral density of the produced sound increased over a wide frequency range as the flow rate and constriction level increased. The overall sound pressure level (OASPL) over several frequency bands was calculated and it was found to be related to the Reynolds number in the smallest cross-sectional area of the constriction. When constriction was less than 50%, the OASPL in the frequency range of 200–800 Hz increased as the Reynolds number increased. In the 75% constriction case, a smaller increase of OASPL was observed. In the frequency range of 150–10 000 Hz, all models demonstrated similar relationships between OASPL and Reynolds number. In the majority of frequency ranges, a Reynolds number of 4000 was required to generate 2 dB OASPL, and OASPL showed dramatic increases with higher Reynolds numbers. To find the source location based on Lighthill’s sound analogy, turbulence strength measurements were performed 5 mm downstream from the constricted area. Small turbulence was observed, indicating that the sound sources were nearby. Our results show that the OASPL increase of the lung sound can be an indicator of the constriction presents in the airway.

Key words: Bronchoconstriction, Respiratory sound, Respiratory auscultation, Adventitious lung sound, Aerodynamic sound

1. Introduction

Listening to the sounds of the lung has long been an important part of lung health diagnosis. It became especially influential after the invention of the stethoscope by Laennec in 1816 (Dalmay et al., 1995). It is an effective and convenient method of detecting lung abnormalities; however, its accuracy in identifying specific diseases is still very low. To achieve better diagnosis accuracy, a more in-depth understanding of sound generation and its alteration mechanisms is required. The sources and the generation mechanisms of lung sound have been addressed previously, particularly after Laennec published a treatise on the diagnosis of lung and heart diseases (Laënnec and Forbes, 1834). Several terminologies were introduced to characterize different sounds based on their pitch and timing (Murphy, 1981; Pasterkamp, 1997), whilst Forgacs (1969) tried to describe possible mechanisms of the sound generation in his report. Hardin and Patterson (1979) proposed generation mechanisms involving the motion of vortices in the human lung. In their hypothesis, monophonic sound would be generated by the air flowing in a certain airway configuration at a certain airflow rate. However, the monophonic sound was not observed in the experimental observations on canine airways.
(Kraman, 1990). Kraman (1990) instead detected a white noise sound profile when air was flowing through the cast canine airway model, at any given measurement point and under any flow rate conditions.

The mechanisms of sound generation in the lung are important in order to define the location of the disease-affected area. This knowledge will also help in determining the symptoms of the diseases and may aid in treatment. In both healthy and diseased lungs, sounds are produced every time we breathe, both in the inspiratory and expiratory directions (Dalmay, et al., 1995). Gavriely et al. (1984) proposed a mechanism of abnormal lung sound generation. That particular study included five theories of wheezing sound production commonly found in asthma patients. These were turbulence-induced wall resonators, turbulence-induced Helmholtz resonators, acoustically stimulated vortex sounds (whistle), vortex-induced wall resonators, and fluid dynamic flutter. The authors suggested that the vortex-induced wall resonator and fluid dynamic flutter were the production methods that agree best with the experimental observations of the wheezing generation, and described an interaction of solid and fluid components in the lung that was able to generate the sound. In the field of aeroacoustics, the sound source is believed to be generated by a fluctuating momentum flux solved in the Reynolds equation (Lighthill, 1952, 1954). From the point of view of plane aeroacoustic theory, the most significant two factors believed to generate the aeroacoustic sound source are examined here in our study; the first, fluctuating jet flow produced by a constriction; the second, the downstream obstacle impinged by the jet.

In an asthma attack, the narrowing of the airways, also known as bronchoconstriction, is commonly observed, and may lead to jet flow production and the increase of flow impingement to the airway junction. This bronchoconstriction is also believed to cause alterations of the highest frequency sound in the inspiration according to a study by Habukawa et al. (2010). In our attempt to study the factors necessary to change the aeroacoustic sounds in the inspiration maneuver as bronchoconstriction appears, we performed an experimental study using a simplified model of an airway junction. The model was a simplified version of those commonly used for analyzing and reproducing the flow pattern in the lung airway (Hardin and Patterson, 1979; Pedley et al., 1970; Schroter and Sudlow, 1969). The effects of the airway narrowing were studied using a constricted tube in the mother branch to represent a bronchoconstriction. Furthermore, the relationship between the sound and the flow fluctuation, which is predicted to be one of the sound generation mechanisms in the constricted branch, is also discussed.

2. Methods and experimental setup

Because of the complexity of human airways, it is common to simplify the model of an airway junction for flow analysis. A model proposed by Pedley (1970) is often used to study lung flow characteristics computationally (Kleinstreuer, Zhang, and Li, 2008; Zhang, Kleinstreuer, Donohue, and Kim, 2005) and experimentally (Adler and Brucker, 2007; Fresconi, Wexler, and Prasad, 2008; Leong, Smith, and Wang, 2009). However, some parameters (e.g., angle of branch and diameter ratio of daughter and parent branch) are considered to have small effects on the velocity profile of the daughter branches and on the sound generation in a junction. Therefore, we further simplified the experimental model used in Pedley (1977) into a T-branch model as a representation of an airway junction to reduce the complexity of model geometry.

In our experimental model, the splitting angle of the mother and daughter branches were 90°, and they were located on the same plane, as shown in Fig. 1(a). All tubes were constructed from silicone material (MegaPinkSil, Megadental GmbH, Büdingen, Germany) while the branching part of the separation point was made of plaster and was three-dimensionally printed (Zprinter 250; 3Dsystems, Rock Hill, South Carolina, USA). An inspiratory maneuver was reproduced by flowing air to the experimental model from the mother branch. The generated sounds were measured using a microphone and the flow fluctuation using a hot wire anemometer inside the tubes.

A normal tube and constricted tube were representative of the airway in normal and bronchoconstriction conditions, respectively. Dimensions of the silicone tube parts are shown in Fig. 1(b) for the normal condition and Fig. 1(c) for the bronchoconstriction case, and the dimensions of the branches are shown in Fig. 1(d). We modeled the bronchoconstriction severity by applying different \( D' \) to the constricted tube model shown in Fig. 1(c), and we define the constriction percentage as follows.

\[
C = 100 \frac{D_0 - D'}{D_0} \%.
\]  

(1)

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Four different constricted tube models were constructed: no constriction model, 25% constriction model, 50% constriction model, and 75% constriction model. All models were constructed with a major diameter ($D_0$) of 7 mm. We considered only one constricted tube to be located in the mother tube, and air was flowed from the mother branch to the daughter branch to represent the inspiratory flow of the respiration. In the branching part (Fig. 1(d)), rounded edges with a radius of 3 mm were used instead of sharp edges in the intersection of two tubes. These rounded edges reduce the flow separation that can be generated by sharp edges, which may have produced unrealistic sound and flow separation.

The setup used for the measurements is shown in Fig. 2(a) and the measurement location of the hot wire anemometer is shown in Fig. 2(b). Air at room temperature, was compressed using an air compressor (Kapsel-con YC-4RS; Holmi, Tokyo, Japan) and sent to an air tank (Anest Iwata SAT-120C-140l Anest Iwata Corp., Yokohama, Japan), until the pressure in the tank reached 1.2 MPa. After reaching this pressure, the compressor was stopped so that noise coming from the compressor engine could be removed. Air was then flowed from the air tank to an expansion box through a rubber tube, whilst passing an airflow meter (Azbil CMS0050; Azbil Corp., Tokyo, Japan) to measure the airflow. The expansion box was made from 15-mm-thick wood plates, and the outer dimensions of the box were 40×38×40 cm$^3$. The inner side of the expansion box was covered by anechoic urethane to reduce the resonance inside the box. A honeycomb channel (Honeycomb V-13-100; Shin Nippon Feather Core Co., Ltd., Saitama, Japan) with a 13-mm cell size and length of 5 cm was located 20 cm from the inlet of the expansion box to generate laminar flow in the outlet of the box. A smooth 3D-printed profile was placed at the outlet to reduce the sound generated by sharp edges. The outlet of the expansion box was connected to the silicone tube of the experimental model.

![Fig. 1](image1.jpg)

(a) T-branch model configurations, parts made of silicone are indicated in pink, while parts made of plaster are indicated in blue. (b) Dimensions of the silicone tube for the normal condition. (c) Dimensions of the silicone tube for the constricted condition. (d) Dimensions of the 3D-printed plaster T-branch; sharp edges at the T-branch were rounded at a 3-mm radius. Dimensions are shown in millimeters.

![Fig. 2](image2.jpg)

(a) Experimental setup used for sound measurements, and (b) hot wire anemometer location for the velocity measurement downstream of the constriction in the mother branch.
A free field microphone (Bruel Kjaer 4939, attached to Bruel Kjaer 2670 pre-amplifier; Brüel Kjær Sound and Vibration Measurement A/S, Nærum, Denmark) was located 8 cm above the test model to record the sound. The sound was conditioned using a Nexus Bruel Kjaer signal conditioner and was sampled in a National Instrument data acquisition NI PXIe-4492 (National Instruments Corp., Austin, Texas, USA) at 20,000 Hz and 16 bit depth. Sound measurements were performed inside an anechoic cover size of 90×90×92 cm$^3$ to reduce the background noise level. The cover was constructed using 10-mm-thick plasterboard, whose inner side was covered by anechoic urethane and whose outer surface was covered by a soundproof sheet (Sandam K-PRO; Zeon Kasei Co., Ltd., Kanagawa, Japan). This anechoic cover was able to reduce 5 dB of the background noise overall sound pressure level (OASPL).

To characterize the sound, the OASPL of the recorded sound across several frequency bands was calculated. The OASPL is a parameter that describes the average pressure level over a specific frequency range, as calculated by the following equation:

$$OASPL = 20 \log_{10} \left( \frac{\sum_{i=a}^{b} P_{ref} \times 10^{PSD(f_i)/20}}{P_{ref}} \right)$$

(2)

where $P_{ref}$ is the reference pressure equal to $2 \times 10^{-5}$ Pa, $PSD(f_i)$ is the PSD value in decibels of the corresponding frequency $f_i$, and $a$ and $b$ are the lower and upper limits of the frequency of interest, respectively. The OASPLs of the sound under all measurement conditions were referenced against the OASPL of background noise, meaning that we obtained a value that was relative to the background noise level.

To understand the flow condition required for the sound generation, several flow conditions were introduced. Airflow rates of 0, 5, 10, 15, 20, 25, and 30 L/min were used and kept constant during measurements. These flow rates were considered valid to represent the quasi-steady state of a breathing maneuver that is commonly found at a certain time span in a breathing cycle. These flow rates correspond to Reynolds numbers (at the inlet of the models) between 0 and 6000, which are Reynolds numbers commonly found in the third generation of a human airway branch. The generated sounds were recorded and analyzed by applying Fast Fourier Transforms (FFT) of 2048 points. An average FFT of 50 data, with an overlapping coefficient of 0.5 and a Hanning window, comprised each data set, and 10 data sets were used to observe the variability of the sampling.

To obtain the aeroacoustic sound source, we measured the flow inside the tube using a single probe, hot wire anemometer (Kanomax 0251R-T5; Kanomax Inc., Osaka, Japan). The hot wire anemometer was located at the center of the tube, 10 mm before the end of the mother tube, as shown in Fig. 2(b). Generally, flow velocity, $u$, at a certain point inside a tube can be characterized as:

$$u = \bar{u} + u'$$

(3)

where $\bar{u}$ is the mean velocity and $u'$ is the fluctuating velocity component. The mean velocity itself can be calculated as

$$\bar{u} = \frac{1}{n} \sum_{i=1}^{n} u_i$$

(4)

where $n$ stands for the number of data used for the analysis. To characterize the flow disturbance, which might be one of the sound sources, we calculated the velocity fluctuation strength (or turbulence strength, TS) using:

$$TS = \sqrt{\frac{1}{n} \sum_{i=1}^{n} (u'_i)^2}$$

(5)

In our measurements, a sampling frequency of 44,000 Hz was used to sample the flow velocity. Each velocity data set contained data of 1.5 s, which corresponds to an $n$ equal to 66,000. This value of $n$ was chosen based on the minimum value required to observe flow fluctuation in the range of 50–10,000 Hz, and was confirmed to be enough to accommodate...
the quasi-periodic phenomenon by the low-frequency fluctuation. For each configuration, we averaged the measurements of 10 data sets to ensure the stability of each measurement.

3. Results
3.1 Aeroacoustic sound

The measurements of the sound were performed at flow rates of 0–30 L/min, but the results shown in this section are only those measured at 5 L/min, 15 L/min, and 25 L/min. Figure 3 shows the power spectral density (PSD) distribution of the sounds for all models at different flow rates. In Fig. 3(a), the results of the no constriction case are shown. At flow rates of 5 L/min and 15 L/min, no sound was generated, but at a flow rate of 25 L/min several small peaks of less than 30 dB appeared at frequencies greater than 300 Hz. Similar results were also obtained in the 25% constriction model shown in Fig. 3(b). However, at a flow rate of 15 L/min, at frequencies between 1300 and 1700 Hz, sound was also measured at levels less than 20 dB. In comparison to the no constriction case (Fig. 3(a)), the increase in PSD was highest at 25 L/min, and the frequency range was also broader. In the 50% constriction case (Fig. 3(c)), sound was not detected at flow rate of 5 L/min, but the PSD level increased sharply to 30 dB when the flow rate was 15 L/min. As the flow reached 25 L/min, broadband sound was detected, reaching a maximum PSD of 35 dB in the frequency range of 100–3000 Hz. Additionally, a valley was observed in the frequency range between 500 and 1300 Hz. For the 75% constriction case (Fig. 3(d)), sound recorded at a flow rate of 5 L/min reached 20 dB at frequencies greater than 1300 Hz. This was not found in the case of the 50% constriction model at the same flow rate. A higher PSD was also recorded for flow rates of 15 L/min and 25 L/min. An increase of PSD of greater than 15 dB was observed when the flow rate was increased from 5 L/min to 15 L/min; however, only small increases of 5 dB were seen when the flow rate increased from 15 L/min to 25 L/min.

![Fig. 3](image-url) The PSD distribution of sounds for all models in four flow conditions. (a) No constriction model, (b) 25% constriction model, (c) 50% constriction model, and (d) 75% constriction model. Flow conditions are shown by different line styles.
3.2 Turbulence strength

Turbulence strength (TS) measurement results are shown in Fig. 4. In all models, the increase of flow rates was followed by an increase of the TS. Each plot shows a different saturation value of the TS and in each model different flow rates were required to reach this saturation value. The log scale of the TS shows a small increase in all cases of less than or equal to 50% constriction. In the no constriction case, a TS saturation value of around 0.1 m/s was observed, which is similar to the value observed in the 25% constriction case. Under 50% constriction, a saturation value of around 0.2 m/s was observed. However, at a constriction level of 75%, the saturation value reached approximately 3 m/s as the flow rate increased. This was the maximum TS value obtained from all models.

4. Discussion

Flow-induced noise and aerodynamic interaction with the surface of a body are known to be two of the sources of lung sound (Gavriely, et al., 1984; Gavriely and Cugell, 1996); however, it is not clear what the most significant factors in sound generation are, or how they contribute to the sound alteration in constricted airways. In the present study, we aimed to uncover the effects of constriction on the jet formations and aeroacoustic sound generations in situations with moderate Reynolds numbers. We quantitatively determined the characteristics of the flow and airway geometry (constriction severity), which are considered to contribute to the sound generation and alteration. We found that the constriction level of the model greatly influenced PSD spectra distribution. Furthermore, each model demonstrated relatively consistent PSD spectra characteristics (as determined by the similar location of the peaks and by the PSD profile) across different flow rates. This clearly shows that the sound frequency distribution of a model strongly depends on airway geometry and that the level of the PSD depends on the air flow rate (which also determines the Reynolds number in the constricted region). These results are contradictory to the theory proposed by Hardin and Patterson (1979), but in agreement with observations of lung sound over a range of flow rates in the inspiratory maneuver (Gavriely and Cugell, 1996; Harper, et al., 2003). In some further studies, it was found that the characteristics of the inspiratory sound (mean, median, or the highest frequency) were changed when the patient performed a methacholine challenge test (Habukawa et al., 2010). The change in the PSD distribution profile of the lung sound could therefore be a clear indicator...
of the severity of the bronchoconstriction in the lung airway. However, a more thorough analysis method is required to predict the bronchoconstriction severity of the complex lung airway by only using the frequency analysis.

OASPLs for the frequency ranges 0–200 Hz, 200–400 Hz, 400–600 Hz, 600–800 Hz, 800–1000 Hz, and 150–10,000 Hz were calculated with respect to the Reynolds numbers obtained from the minimum cross-sectional area of the constriction, and are shown in Fig. 5. From these figures we observe that the OASPL of each model did not increase in the frequency range of 0–200 Hz, even though the Reynolds number had reached its maximum observable value. This frequency range was commonly disturbed by background-noise, and the steady OASPL value is a clear indicator of the constant noise that appeared in all measurements. Additionally, sound generated by vibrations of solid elastic material, which is commonly found in this frequency range, was not found in our experiments. This indicates that the elastic solid vibration sound was not generated in our experiments, or generated at such a low level that it could be neglected.

In the frequency ranges of 200–400 Hz, 400–600 Hz, 600–800 Hz, and 800–1000 Hz the sound starts to appear when the Reynolds number reaches 4000, with an average increase value of 2 dB. In the frequency range of 200–1000 Hz, we observed that models with constriction less than 50% produced higher OASPL when compared with models with 75% constriction, for a Reynolds number greater than 5000. In Fig. 5(f), we observe the same trend as other constriction cases, whereby the OASPL increases with Reynolds number. This is an indicator that the 75% constriction model generated higher frequency components when compared with the other models.

The increase at low frequencies in low-constriction models can be related to the flow velocity and the geometry of the models. A low-constriction level means that the diameter is still relatively wide at the constricted region. This large diameter corresponds to a lower-frequency region. A large diameter leads to a low-flow velocity and low-vorticity speed, which may generate lower-frequency sound. Higher constriction levels lead to an increase in the flow velocity and therefore more energy being contained in the flow, which in turn generates the flow instability that is one of the sources of the sound. The combination of high flow velocity and the constriction geometry result in an increase in the 200–800 Hz frequency range. In the higher constriction level model, compressibility of the flow must also be considered and this may lead to a significant increase in the higher frequency range. Physically, high frequency sound might be reduced during lung sound measurement from the chest surface because lung parenchyma acts as a low-pass filter, with a general threshold frequency of 1000 Hz (Schreur et al., 1995), or 2000 Hz (Harper et al., 2003).

Combining the TS measurements (shown in Fig. 3) into a single plot, we obtained a relationship between TS and Reynolds number, as presented in Fig. 6. From the TS measurements, we found that when measured in the center of the tube, 5 mm downstream from the minimum constricted diameter, the flow showed a small fluctuation in the cases of no constriction, 25% constriction, and 50% constriction. This means that in the vicinity of the measurements, the flow is relatively stable (i.e., not fluctuating or with only small turbulence) in the case of less than 50% constriction, even though the Reynolds number was as high as 14,000. In the case of 75% constriction, the flow becomes unstable and the fluctuation level is proportional to the Reynolds number of the flow. However, if we look closely at the no constriction case at 15 L/min, a discrepancy appears and a sudden increase of the TS can be found, which also appears at 75% constriction under a 4 L/min flow rate. This discrepancy may be due to interference from the measurement instrument, which may affect the flow inside the silicone tube and result in high flow fluctuation around the hot wire anemometer.

Flow fluctuation is one of the known mechanisms of lung sound generation, as proposed by Lighthill (1952). Additionally, constriction has been demonstrated to be a source of flow fluctuation (Krane, 2005). The fluctuation of the velocity component \( \mathbf{v} \) in the Lighthill stress tensor \( \mathbf{T} = \rho_0 \mathbf{v} \mathbf{v} \) appears as one of the sound sources because for this fluctuation will be the source of the disturbance of the pressure or the density of the fluid in the Lighthill equation. The velocity fluctuation, indicated by the turbulence strength term, may therefore indicate the aeroacoustic sound source. The geometry of the constriction was responsible for generating a high-flow velocity and high-velocity fluctuations, which were seen when the channel diameter was reduced. This mechanism may explain increased sound in patients with bronchoconstriction. Comparing the flow fluctuation analysis and the OASPL analysis shown in Figs. 5 and 6, we saw that the OASPL did not follow similar behavior to the TS; however, the OASPL has a defined relationship to the Reynolds number of the flow. The increase of the OASPL in the no constriction, 25% constriction, and 50% constriction models was not followed by an increase of the flow fluctuation. In our system, the flow fluctuation appeared in some regions inside the models, but its exact location has not yet been detected. The measurements taken from a location downstream from the constriction did not agree with the OASPL data.
Fig. 5 OASPL of the sound measured from the T-branch model. (a) 0–200 Hz, (b) 200–400 Hz, (c) 400–600 Hz, (d) 600–800 Hz, (e) 800–1000 Hz, (f) 150–10,000 Hz.
In a real human lung, bronchoconstriction is known to change the properties of the lung sound (Habukawa et al., 2010; Schreur et al., 1995; Spence et al., 1992), although specific parameters that can objectively predict the bronchoconstriction level are still under investigation. Bronchoconstriction can appear at several locations in peripheral airways simultaneously. This situation leads to an increased number of sound sources, and therefore smaller Reynolds numbers are required for the sound detection, but more complex frequency distributions are generated. From our observations, we suggest that this increase in sound sources will affect the medium frequency component of 200–800 Hz for constrictions of less than 50%. In the normal condition, assuming a high respiratory flow rate (100 L/min), Reynolds numbers of greater than 4000 might appear only up to the third generation of the airway (Pedley et al., 1970). In a diseased lung, this Reynolds number might appear in higher generations because of the appearance of constriction in the lung airway, leading to a spread of turbulence sound source locations.

5. Summary

We performed an experimental study to understand the factors affecting sound generation and alteration in a bronchoconstriction. Using T-branch models as a representation of airway branches, and a constricted tube as a representation of airway bronchoconstriction, we observed changes in sound PSD caused by changes in flow rates and changes in frequency distribution produced by the constriction level. The combination of the flow rates and the constriction changes the Reynolds number of the flow, and the constriction level itself determines the sound characteristics. Constriction levels of less than 50% tended to increase the OASPL within a frequency range of 200–800 Hz, while constriction levels of 75% tended to increase OASPL at frequencies greater than 800 Hz. A minimum Reynolds number of 4000 was needed to generate sound with OASPL of 2 dB in our experiments. At Reynolds numbers lower than this, the OASPL was relatively small, and at higher Reynolds numbers the OASPL increased drastically. The constriction is known to increase both the Reynolds number of the flow and the OASPL results; however, flow fluctuation measurements at the center of the tube, 5 mm downstream of the constriction, did not show the same trends. Lighthill’s theory predicts this location to be a sound source, but our results suggest that the flow fluctuation might appear not only at the assigned measurement point, but also somewhere else further downstream from the constriction. Additionally, we observed that a constriction in the lung airway may increase the frequency distribution of the generated sound. A clear indicator of the constriction is the increase of the OASPL generated by the high Reynolds number flow, which is generated by a high flow rate breathing maneuver. This increase of OASPL, if measured from the human chest, could therefore be useful in detecting the existence and severity of a constriction. In the future, studies using more physiologically accurate models could ascertain the validity of our current analysis for in-vivo situations.

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