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The measurement of airflow using Singing helmet that allows free movement of the jaw

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Abstract

Objectives—Airflow measurement is a useful method of evaluating laryngeal physiology. We introduce a non-invasive device that measures airflow without restricting jaw movement or requiring phonation into a mouthpiece, thus facilitating measurement during singing and connected speech.

Study Design—Validation and human subject trials were conducted. Airflow measurements were obtained from sixteen male and sixteen female subjects during singing, speech and constant vowel production tasks.

Methods—A similar helmet was designed by Stevens and Mead in 1968. The new device validity was evaluated by comparing the measured volume of air to a known volume of administered air using a calibration syringe. Subjects were asked to voice sustained vowels at low, medium and high vocal intensity, read two sentences at a conversational volume, and perform different singing exercises while airflow was recorded.

Results—The device accurately and reliably measured airflow with mean airflow values falling within previously published ranges. There was an experimentally determined response time of 0.173 ± 0.014 s. Subjects were able to comfortably perform speech and singing exercises. Male subjects had higher airflow for all sustained vowels ($p < 0.05$). Airflow was higher for abduction rather than adduction sentences ($p < 0.05$).

Conclusions—No other portable device has been shown to measure airflow during singing and speech while allowing for free movement of the jaw. This device provides a more natural environment to measure airflow that could be used to help evaluate laryngeal function and aid in singing training.

Keywords

Airflow measurement; singing; aerodynamics; connected speech

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INTRODUCTION

Airflow measurement is a useful and noninvasive method of evaluating laryngeal physiology. Laryngeal pathologies, functional voice disorders, including vocal nodules, polyps, and neuromuscular disorders, are often characterized by abnormal airflow or variable flow rates resulting from incomplete closure or hyperadduction of the vocal folds during sustained phonation^{2,3}. For this reason, airflow measurement is used to assess both vocal disorders⁴ and some neuromuscular disorders of the larynx and velum⁵. Such measurements can objectively and noninvasively assist in initial diagnosis and continual monitoring of disease progression and treatment⁶.

Airflow is traditionally evaluated noninvasively while subjects produce and hold a vowel at a constant amplitude and frequency in order to produce a stable airflow. Clearly, however, flow measurements during sustained vowels are not reflective of airflow profiles for connected speech, as no transitions between sounds are required^{7,8}. As a result, the development of devices and methods capable of capturing airflow during speech has been of great interest to the field. One of the earliest attempts was made by Klatt et al. by using a facemask with a 4 cm diameter hole covered by a fine mesh screen serving as a resistor¹⁰. A pressure transducer was then used to measure the pressure within the mask to allow for calculation of airflow. While this effort was successful in measuring flow during speech production, the mask restricted movement of the jaw and reduced auditory feedback¹⁰.

Similarly, Hixon introduced a slightly improved prototype of the mask that included an airtight seal with the subject's face while still providing some jaw movement¹¹. Nevertheless, the device did not offer completely free movement of the jaw. Stone et al. attempted to measure the airflow of subjects while singing, but used a Rothenberg mask that did not allow for free movement of the jaw¹². Alternatively, Hixon also utilized a modified body plethysmograph to measure transglottal flow¹¹. In this technique, a dome was placed over the head of the subject and an airtight seal was made with the neck. Though this design provided completely free jaw movement, the plethysmograph is not portable.

Current airflow measurement is thus hindered by available technology. In addition to the limits placed on speech analysis by current techniques, to our knowledge none have adequately addressed the possibility of airflow analysis during singing. Although speech and singing are similar in many ways, singing may require greater frequency ranges, effort¹³ and jaw movement¹⁴. Consequently, a new approach is needed to enhance the ability to measure airflow during speech as well as singing. The goal of this study was to develop a device for measuring flow during singing and speech in a manner that does not inhibit the free movement of the jaw or distort higher frequency sounds. Such a device would allow for improved analyses of both general and professional voice users. We hypothesize that the helmet will reliably measure airflow values.

MATERIALS AND METHODS

Device

A schematic of the device is shown in Figure 1A. A $30.5 \times 30.5 \times 30.5$ cm plastic cube created a 3.7 kg sealed helmet (Figure 1B). The volume in the helmet was measured to be 0.028m^3 without a participant and 0.272m^3 with a female participant and 0.271m^3 with a male participant. The helmet was made of clear acrylic sheets joined by strong acrylic cement (Weld-On Acrylics, IPS Corp., Gardena, CA). A custom acrylic sheet with a 22.9 cm diameter hole (Acrylic Design Works, Chicago, IL) was used on the inferior face, creating an opening for the subject's head. A seal was created between the subject's neck and helmet using a latex dry suit (Bare, Langley, British Columbia). The neck seal attached firmly to the helmet around the 22.9 cm hole and tapered to fit snugly around the patient's neck. Airflow holes with diameter of 5.1 cm were drilled on the anterior and both lateral faces of the cube. Two layers of synthetic felt were placed in the orifice of the airflow holes and were secured to the helmet to provide a known resistance to expired air, creating a linear pressure drop across the surface. The fabric was a synthetic felt which showed a linear pressure drop. A differential pressure transducer (DC001NDR5 Sursense DC pressure sensor; Honeywell Inc., Morristown, NJ) was housed on the superior face of the cube with one port connected to the helmet and one open to the surroundings (atmospheric conditions in Madison, WI). A digital sound level meter (2200 SLM; 3M, Oconomowoc, WI) placed about 61 cm, a distance deemed more comfortable for testing situations, in front of the subject provided real-time visual sound pressure level (SPL) feedback and acquired an acoustic trace. Ambient noise was controlled for by testing subjects in a sound-proof booth. Output voltages from the pressure transducer and sound level meter were gathered using a National Instruments data acquisition board (model NI USB-6218 BNC; National Instruments Corp, Austin, TX) and recorded using customized LabVIEW 8.5 software (National Instruments Corp). Pressure data were recorded at 100 Hz and acoustic data at 40,000 Hz.

System Calibration

An aerodynamic analog of Ohm's Law has been extensively applied to model laryngeal aerodynamics:

$$P = U * R$$

where P is pressure, U is airflow, and R is the resistance of the pathway. This relationship formed the basis for calculating flow from pressure measurements using a standard curve. The general theory behind calibration was to vary airflow with a pressurized air source, measure corresponding pressure changes with the differential pressure transducer, and calculate the rate of change between this relationship.

A customized LabVIEW 8.5 program gathered the corresponding voltage output from the pressure transducer at flow rates adjusted in increasing increments of 0.05 L/sec from 0.00 to 0.75 L/sec as measured by an Omega airflow meter (model FMA-1601A; Omega Engineering Inc, Stamford, CT). The resultant pressure-flow relationship was then plotted for post-processing. System gain and offset were integrated into the aforementioned LabVIEW data collection program. The calibration was completed by calculating resistance

from the fabric by means of a least square linear fit from the plotted relationship. Airflow could then be determined from the Ohm's law analog by solving for airflow given the calibrated resistance and changing pressure values in the helmet. Calibration was performed immediately before every subject to account for small perturbations in room airflow ventilation or atmospheric conditions, and was done with the neck-hole sealed.

Device Validation and Quantification

The accuracy of the device was assessed by administering a known volume of air into the helmet and comparing that to the volume measured by the instrument. A 1 ± 0.012 L Viasys calibration syringe (model 720252; CareFusion Corp, San Diego, CA) ejected air at a slow, controlled speed so the air could be modeled as an incompressible fluid. Forty-five traces generated by the helmet were evaluated. The area under the airflow trace was integrated to yield total volume ejected during each trial.

The device was expected to exhibit a time-weighted response to changes in airflow due to the overall volume of the device and the natural compressibility of air. As noted by Rothenberg, modeling the measurement system with an ideal circuit analog to determine the mask response time becomes highly complex if the difference between the mouth and the walls of the mask (helmet) are not "small compared to the vocal tract"¹⁵. Thus, the response time was determined experimentally. A near instantaneous influx of air at approximately 0.417 L/sec (20 L/min) was administered to the helmet to measure the response time for 10 trials. The response time of the device was defined as the length of time for the device to fully equilibrate to the abrupt increase in airflow. This response time is approximately equal to the 5th time constant, τ , defined as the time at which the signal reaches 99.3% of the final value. The highest pressure that we recorded when assessing the leakage of the neck seal was 12.4 cm H₂O. This was not the maximum pressure before leakage occurred as we did not raise the pressure any higher, but well above any realistic pressure values that could be inside of the system during participant testing. The testing of neck seal tightness was conducted by simply sealing the openings for our fabric resistors and increasing airflow supply in the closed system from the same air supply used in calibration. We ceased this operation at 12 cm H₂O in conclusion that the neck seal will refrain from leaking under even the highest operating pressures (2 cm H₂O).

Human subject testing

Approval for human subject recruitment and testing was obtained from the Institutional Review Board (IRB) of the University of Wisconsin-Madison prior to the experiment. Thirty-two subjects were recruited from the university campus to participate in this study. We recruited 32 subjects to achieve a standard power level of 0.8 based on previously published mean male and female airflow values^{16,17}. Any individual over the age of 18 without a history of respiratory or laryngeal disease, latex allergy, and/or claustrophobia was eligible to participate. Subjects were excluded if they indicated a history of smoking as these subjects tend to have a lower peak expiratory airflow²². 16 Male and 16 Female subjects were recruited and were 21.16 ± 1.80 years old.

Following calibration, the subject's head was placed inside the helmet and sealed with the latex neck seal. A constant (DC) bias airflow of 25 ± 2 L/min was filtered and passed into the helmet for subject comfort by flushing the dead space with fresh air. Applying the superposition principle for linear systems, the bias flow values were subtracted from the airflow measurements during data analysis. The helmet was padded and rested on the subject's shoulders. Subjects were asked to inform the experimenter if the bias flow was insufficient or the helmet was uncomfortable.

Exercises

Subjects were asked to voice continuous vowels / α /, / e /, / i /, / o /, / u / at 60 ± 2 , 65 ± 2 , and 70 ± 2 dB to observe changes in airflow among different vowels and vocal intensities. These intensities correspond to low, medium and loud conversational volumes. Recordings were not taken until the subject was in the specified range, as indicated visually by the SPL meter. Each token was repeated five times and in the same order for all participants.

Subjects were also asked to read sentences at a normal conversational volume. Two sentences were selected based on their use in evaluating patients with spasmodic dysphonia (SD). The first, "We eat eels every day" is used to elicit vocal fold adduction, and leads to lower airflow. The second, "He has half a head of hair" is used to elicit abduction, and generates higher airflow. Each sentence was repeated five times per subject. Airflow was calculated for each task and averaged for each subject.

Finally, subjects were asked to sing. A warm-up scale, starting at middle-C was used to assess the airflow rates at different fundamental frequencies. An audio sample of middle-C was provided to each subject. Intensity and frequency were not monitored for this task. Subjects then sang "Row, Row, Row Your Boat." to show its efficacy in providing airflow values during song. The scale and "row" tasks were conducted once per subject. Subjects ranged from no singing background to medium singing background. These tasks were used to qualitatively demonstrate the helmet's ability to record airflow changes during more complicated tasks.

Data and Statistical Analysis

Data were imported from the LabVIEW acquisition program and analyzed using custom MATLAB programs (The Mathworks, Natick, MA). The onset and offset of phonation was determined using the acoustic signal. These time points were then applied to the airflow data to analyze data only during phonation.

Paired T-tests were used to evaluate differences in subject airflow between soft/loud vowel production and adductor/abductor sentences tasks. Potential differences as a result of sex were evaluated using independent t-tests. In all cases, the Shapiro-Wilk and Equal Variance tests were used to determine whether parametric testing could be applied. A Two-Way Repeated Measures analysis of variance (ANOVA) was performed to test for differences and interactions between sex and ad/abductor sentence airflow values. If significant differences were found, pairwise comparisons were made via the Holm-Sidak method. To test for differences among vowels, and among the three sound pressure levels, a two-way repeated measures ANOVA were performed. Significance was determined using $\alpha = 0.05$. All

statistical analyses were completed using *SigmaPlot 11.0* software (Systat Software, Inc., Chicago, IL).

RESULTS

Validation of device

The results from 45 validation trials are shown in Figure 2. The measured mean ($\pm$ standard deviation) volume of air that was passed through the device was $1.016 \pm .001$ L and approximated that of the known air volume ($1.0 \pm .012$ L). The step response time over 10 trials was 0.173 ± 0.014 s.

Subject exercises

Under sustained vowel phonation at a given sound pressure level, differences in airflow between vowels were seen within each individual subject with high standard deviations for each level. Across the group, no significant difference was found among different vowels. In addition, subjects did not exhibit significant differences in airflow among sound pressure levels for a given vowel. The cross-gender comparison found that males had significantly ($p < 0.05$) greater airflow on all vowel tasks (Table 1). Comprehensive statistical data for this comparison can be seen in Table 1. Averaging over all vowel tasks, males were found to have a mean ($\pm$ standard deviation) airflow of 0.193 ± 0.101 L/sec at low intensity, 0.202 ± 0.104 L/sec at medium intensity, and 0.213 ± 0.109 L/sec at high intensity. Females exhibited an average airflow of 0.116 ± 0.0604 L/sec at low intensity, 0.122 ± 0.0618 L/sec at medium intensity, and 0.123 ± 0.0685 L/sec at high intensity. Figure 3 shows the summary data for the two sexes.

For the sentence tasks, two-way repeated measures ANOVA showed a significant difference between the sexes $F(1,30)=31.28$, $p<0.001$ and between the sentences $F(1,30)=227.79$, $p<0.001$. As expected, the abductor sentence produced greater airflow amplitudes than the adductor sentence. In addition, there was a significant interaction between sex and ab/adductor sentences $F(1,30)=22.70$, $p<0.001$. Pairwise comparisons via the Holm-Sidak method for the sentence tasks are reported in Table 2. Further pairwise comparisons showed significant differences between males and females stratified by ab/adductor groups, as well as significant differences between ab/adductor sentences when stratified by sex. Data collected for the scale and song singing tasks can be seen in Figure 4. Large dips in airflow correspond to inspirations. An example of a "Row, Row your boat" airflow/audio trace is shown in Figure 4A and 4B while a singing scale example is displayed in Figure 4C and Figure 4D. The device was shown to accurately measure airflow in dynamic conditions and at different frequencies.

DISCUSSION

The validation trials demonstrated that the device is capable of accurately and consistently measuring airflow. Measured volumes differed by an average of +1.6% over 45 trials and the standard deviation was within the range expected from the air source used. The slightly larger value measured by the device could have been due to the imperfect assumption of incompressibility of the fluid and the acceleration of the air when it was administered. The

slow response time of the helmet, when compared to the Rothenberg mask, was expected due to the large volume of the device. The system's slow response defined the minimum time required for equilibration of the system to a step DC airflow before measurements could be taken. Consequently, the time delay defines the minimum time required (173 ms) before accurate measurements of sustained vowel airflow can be taken when starting from a baseline. The delay therefore allows reasonably accurate characterization of airflow for exercises in which large airflow changes occur at a rate of 5Hz or less.

For sustained vowel phonation, the device found no significant differences in airflow between vowels across the population sample. Previous studies suggest average group airflow values between 0.07 and 0.20 L/sec¹⁶, a male range of 0.1–0.3 L/sec, and a female range of 0.09–0.21 L/sec¹⁷. The helmet consistently measured airflows for within these ranges, exhibit large deviations in airflow values^{16,18}, and show elevated airflow rates in male subjects¹⁷, all in agreement with prior studies. Though the difference in mean airflow rate between genders was greater than that observed in previous experiments, this may be explained by the use of a lower range of vocal intensities during tasks. In this experiment, the subjects were asked to obtain the specified intensity ranges while the amplitude and fundamental frequencies were not controlled. A more reliable measurement of airflow during sustained vowel phonation should control these parameters, which—unlike the situation in this study—would be done most easily with experienced vocalists.

Although the measurement of airflow during sustained vowels is useful, the main purpose of the device is to obtain airflow measurements during connected speech and singing. The mean airflow of the abductor sentence was greater than that of the adductor sentence across the sampled population as expected. This indicated the device was successfully able to measure airflow values during normal speech and could identify cases of increased airflow from targeted sentences without interference from the measuring device. This could be useful in the aerodynamic evaluation of disorders most symptomatic during connected speech, such as spasmodic dysphonia.

Finally, we used the device to measure airflow while voicing a scale and singing a song. Figure 4 demonstrates the ability of the system to measure airflow under dynamic conditions despite changes in frequency. When analyzing the airflow signals, noise resulting from electrical interference or from vocal instability can be observed. This could be corrected in future studies by implementing a high-pass filter. From the flow traces, it can be seen that the device captured and recovered from deep inhalations, suggesting that it operated effectively to include large airflow fluctuations found in typical speech and singing. These fluctuations, however, prevented immediate recovery of airflow measurements due to the time delay inherent in the box as well as exhalation. As previously noted, the size of the helmet dictates the rate at which large airflow changes (e.g. changing notes) are discernible, and future applications should allow subjects to hold notes for at least 0.2 seconds to obtain a reliable value. If used in concert with magnetometers that have been shown to follow the displacement of the chest and abdomen²⁰, singing coaches could obtain a more quantitative profile of the respiratory and phonatory mechanics during singing.

The helmet has promise for future applications in voice assessment. Several current limitations will be the subject of future research. First, the device should be compared to currently available techniques. Additionally, the device will be used in future studies to determine the utility of airflow measurements during singing which remain unexamined, as well as compare pathological and normal vocal conditions. Additionally, several subjects commented on the weight placed on their shoulders and the loudness inside the device. The loudness inside the device also likely induces a Lombard effect thus altering the sound production of the participants. Modifications should be made to reduce the weight and decrease reverberation within the helmet. Further studies should seek to quantify the effect that these limitations have on subjects during phonation tasks. Finally, the SPL audio device was placed in a suboptimal position for recording subject phonations. As a result, the sound level and acoustic data recorded do not precisely represent the environment inside the helmet. For future studies that require more accurate measurements of acoustic data, audio should be recorded inside the helmet.

Future studies with the singing helmet will compare pathologic to control groups. Measuring airflow of subjects with vocal nodules of varying size during a variety of vocal tasks, for example, may be useful, as would evaluating patients with SD during different connected speech tasks. This could augment the approach described by Hoffman et al.²¹ using aerodynamic parameters to distinguish adductor and abductor spasmodic dysphonia.

CONCLUSION

A new method to measure airflow during singing and speech was proposed. This novel device uses a known linear relationship between airflow and the pressure drop across a known resistance. The device is portable and allows for free movement of the jaw. This device has potential clinical significance, as it provides a more natural environment to collect airflow measurements that could be used to help evaluate laryngeal function.

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Figure 1.

A) Schematic of the airflow measurement system used in this study. B) A closer look at the airflow helmet device. The subject's head fits through the elastic seal on the bottom and red resistors allow air movement through the device. A biased flow air current is also introduced into the device.

Figure 2.

Results from validating the device by passing a known 1 L volume of air through. The airflow measured was integrated to yield volume from 45 trials. The average (1.016) over all trials is shown as the dotted line.

Figure 3. Airflow during continuous vowel production exercises for males (gray) and females (white). Boxes represent the median and inter-quartile ranges and whiskers represent the 10th and 90th percentile ranges. L: Low intensity phonations M: Medium intensity phonations H: High intensity phonations

Figure 4.

A) Airflow measurement from a subject performing the row exercise. B) Sound pressure wave measured simultaneously to the airflow. C) Airflow measurement from a subject performing scale exercise D) Corresponding sound pressure wave for scale exercise

Table 1

Comparison between male and female airflow values for all vowel phonation tasks. Degrees of freedom and t-value are given for t tests. U statistic is given for Mann-Whitney Rank Sum tests. Male airflow values were significantly higher ($p < 0.05$) in all tokens.

Task	Vocal Intensity	Male Airflow (L/sec)		Female Airflow (L/sec)		t (df = 30)	U	p
/a/	low	Mean	0.177	Mean	0.101	0.0517	-	55 0.006
	med	STD	0.193	STD	0.107	0.0513	3.169	- 0.004
	high		0.206		0.104	0.0647	3.170	- 0.004
/e/	low	Mean	0.190	Mean	0.116	0.0581	-	68 0.025
	med	STD	0.188	STD	0.119	0.0593	-	67 0.023
	high		0.192		0.104	0.0734	-	62 0.014
/i/	low	Mean	0.176	Mean	0.105	0.0588	-	62 0.014
	med	STD	0.190	STD	0.107	0.0530	-	55 0.006
	high		0.201		0.110	0.0647	-	48 0.003
/o/	low	Mean	0.205	Mean	0.124	0.0595	-	52 0.004
	med	STD	0.215	STD	0.137	0.0675	-	61 0.012
	high		0.221		0.132	0.0622	-	62 0.014
/u/	low	Mean	0.217	Mean	0.136	0.0729	-	60 0.011
	med	STD	0.224	STD	0.140	0.0744	-	69 0.027
	high		0.244		0.142	0.0787	-	54 0.006

Table 2

Comparison of abductor (“he”) and adductor (“we”) exercises between sexes and between males and females stratified by ab/adductor groups, as well as between ab/adductor sentences when stratified by sex.

Comparison	Mean and SD for Comparison 1	Mean and SD for Comparison 2	T Value	P Value
Male vs. Female	0.486±0.320	0.228±0.156	5.593	<0.001
Abductor (AB) vs. Adductor (AD) Sentence Airflow Values	0.549±0.268	0.164±0.114	15.093	<0.001
AB vs. AD airflow values within males	0.739±0.246	0.233±0.122	14.041	<0.001
AB vs. AD airflow values within females	0.359±0.107	0.096±0.042	7.303	<0.001
Male vs. Female within AB sentence	0.739±0.246	0.359±0.107	7.199	<0.001
Male vs. Female within AD sentence	0.233±0.122	0.096±0.042	2.593	0.013