The Measurement of Airflow Using Singing Helmet That Allows Free Movement of the Jaw

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Summary: Objectives. Airflow measurement is a useful method of evaluating laryngeal physiology. We introduce a noninvasive 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 16 male and 16 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 seconds. 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.

Key Words: Airflow measurement–Singing–Aerodynamics–Connected speech.

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.[1,2] For this reason, airflow measurement is used to assess both vocal disorders[3] and some neuromuscular disorders of the larynx and velum.[4] Such measurements can objectively and noninvasively assist in initial diagnosis and continual monitoring of disease progression and treatment.[5]

Airflow is traditionally evaluated noninvasively although subjects produce and hold a vowel at a constant amplitude and frequency 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.[6,7] 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[8] by using a face mask with a 4-cm diameter hole covered by a fine mesh screen serving as a resistor. A pressure transducer was then used to measure the pressure within the mask to allow for calculation of airflow. Although this effort was successful in measuring flow during speech production, the mask restricted movement of the jaw and reduced auditory feedback.[8]

Similarly, Hixon[9] introduced a slightly improved prototype of the mask that included an airtight seal with the subject's face while still providing some jaw movement. Nevertheless, the device did not offer completely free movement of the jaw. Stone et al[10] attempted to measure the airflow of subjects while singing but used a Rothenberg mask that did not allow for free movement of the jaw. Alternatively, Hixon also used a modified body plethysmograph to measure transglottal flow.[9] In this technique, a dome was placed over the head of the subject, and an airtight seal was made with the neck. Although 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, effort,[11] and jaw movement.[12] 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 0.028 m^3 without a participant and 0.272 m^3 with a female

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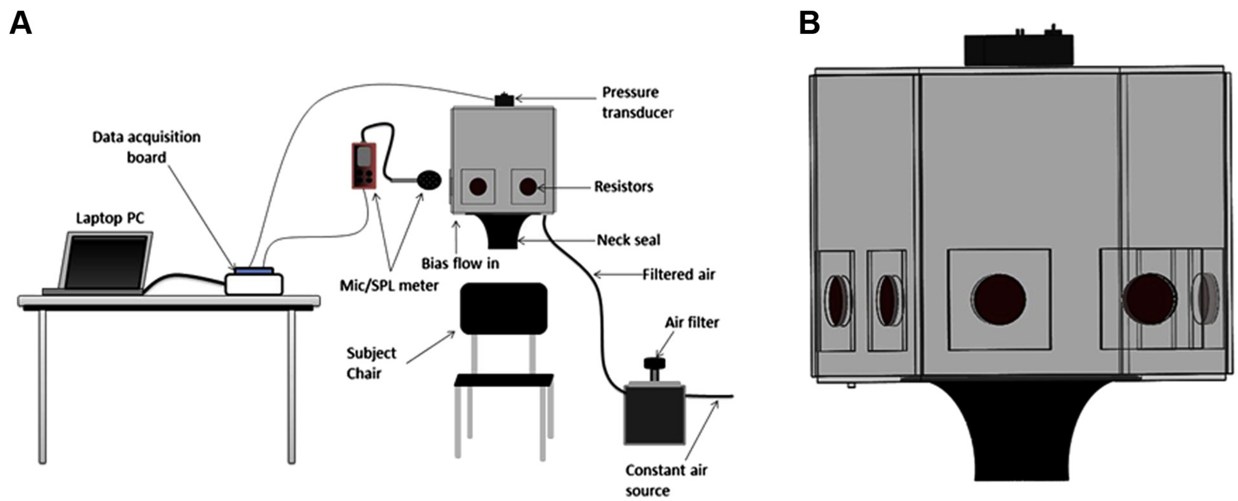

FIGURE 1. (A) Schematic of the airflow measurement system used in this study. (B) A closer look at the airflow helmet device. The subject's head fits through the elastic seal on the bottom and *red* resistors allow air movement through the device. A biased flow air current is also introduced into the device. (For interpretation of the references to color in this figure legend, the reader is referred to the Web version of this article.)

participant and 0.271 m^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 soundproof 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:

$$P = U \times R,$$

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 these relationships.

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/s from 0.00 to 0.75 L/s 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 neckhole 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.