Modeled Velopharyngeal Orifice Area Prediction During Simulated Stop Consonant Production in the Presence of Increased Nasal Airway Resistance

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This project examined modeled velopharyngeal orifice area estimation under conditions simulating voiceless stop consonant production in the presence of nasal airway obstruction. The results indicated that accurate estimates of velopharyngeal orifice area can be obtained using Warren's hydrokinetic equation during aerodynamic events like those known to exist during speech in the presence of increased nasal airway resistance. These findings provide support for clinical and research use of Warren's pressure-flow approach to investigate velopharyngeal function during speech production.

In recent papers, information about the accuracy of estimating modeled velopharyngeal orifice areas using Warren's pressure-flow approach (Warren and DuBois, 1964) was provided. In these studies, estimations were obtained under steady and dynamic airflow conditions (Smith and Weinberg, 1980, 1982, 1983). Results indicated that accurate velopharyngeal area estimations could be obtained under steady airflow conditions (4 to 6% overall error in estimation) and under dynamic airflow conditions simulating the production of voiceless stop consonants (6% overall error in estimation) and voiceless fricative consonants (8% overall error in estimation).

Previous research has shown that individuals with cleft lip and palate have higher nasal airway resistance than normal subjects (Warren et al, 1969). Therefore, modeled velopharyngeal orifice area estimates under conditions simulating extreme degrees of nasal airway obstruction were obtained (Smith et al, 1984). Under these conditions, airflow through the model was not varied. Results indicated that accurate estimates of velopharyngeal orifice area (approximately 3 to 5% overall error in estimation) could be obtained under conditions of extreme nostril obstruction when airflow rates were nonvariant.

During speech production, airflow rates (i.e., pressure-flow events) vary dynamically. There is currently no information about the accuracy of velopharyngeal orifice area estimation obtained in the presence of increased nasal airway resistance when airflow rates vary. Therefore, the purpose of the present project was to quantify the predictive nature of modeled velopharyngeal orifice area calculations in the presence of increased nasal airway resistance under conditions simulating voiceless stop production.

METHOD

Modeling Apparatus

The vocal tract model used in this project was constructed according to the di-

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mensions outlined by Warren and Devereux (1966) and was like that used in our previous modeling studies.[1] In the model, the oral and pharyngeal dimensions of an adult vocal tract are approximated and the cross-sectional area of the model nose offers resistance to airflow comparable to established values for normal individuals.

In this project, velopharyngeal orifice area estimates were obtained under four conditions. In condition #1 (normal condition), the nasal passages of the model were fully open. In conditions #2 to #4, three magnitudes of nasal obstruction were simulated by inserting plugs (tubing) into the model nostrils. The airway resistances offered by these plugs were 9 cm H_2O per liter per second (LPS), 27 cm H_2O/LPS, and 89 cm H_2O/LPS at airflow rates of approximately 0.2 LPS. In condition #2, the left nostril was open or unoccluded while the 27 cm H_2O/LPS plug was inserted into the right nostril, simulating uninasal obstruction. In condition #3, the 9 cm H_2O/LPS plug was inserted into the left nostril, and the 27 cm H_2O/LPS plug was inserted into the right nostril, simulating one condition of binasal obstruction. In condition #4, the 89 cm H_2O/LPS plug was inserted into the left nostril, and the 27 cm H_2O/LPS plug was inserted into the right nostril, simulating an extreme condition of binasal obstruction. Resistances were calculated using the formula $R = P/\dot{V}$, where P is the pressure drop across the plug and V is the volume rate of airflow through the plug (Butler, 1960).

The area of the model velopharyngeal orifice was varied by inserting cover plates over the fully open velopharyngeal port. In this study, six cover plates were used. There was a circular opening in each cover plate to provide known velopharyngeal port areas of approximately 3.10 mm^2, 6.96 mm^2, 12.55 mm^2, 19.72 mm^2, 28.39 mm^2, and 38.12 mm^2. Using a micrometer, these circular areas were calculated from the diameter of the bores used to create the openings. These orifice openings were chosen to sample a wide range of portal openings. The oral port of the model was closed throughout this investigation to stimulate conditions known to exist during stop consonant production.

[1]Vocal tract model design altered by Jerald B. Moon and Bernd Weinberg, Department of Audiology and Speech Sciences, Purdue University. Model constructed by Alfred Nelken and George Mertz, Research Resources Center, Instrument Shop Facility, University of Illinois at Chicago.

Airflow and Differential Pressure Measurements

The volume rate of airflow ($\dot{V}$) through the model was sensed by a Fleisch #2 pneumotachograph. This device was coupled to the right nostril of the model. The pressure differential across the screen of the pneumotachograph was sensed by a Statham PM 15 E differential pressure transducer. Air flow measurements were calibrated to provide full-scale deflection of 519 cc per second. The airflow measurement system was calibrated with a flowmeter (Fisher & Porter, 10A1027).

The pressure differential across the velopharyngeal orifice was transmitted directly to a differential pressure transducer (Statham PM 6) using two catheters. One catheter was coupled to the left nostril of the model, and the second catheter was inserted into the oral floor of the model. Pressure measurements were calibrated to provide full-scale deflection of 10 cm H_2O. A water manometer was used to calibrate these pressure measurements.

Transduced pressure-flow signals were amplified (Grass 7PIF, 7DAF) and digitized through a 12-bit analog-to-digital converter (Digital MNCAA) at an effective rate of 100 samples per second. Converted data were displayed graphically, permitting selection of paired pressure-flow data points by means of a movable cursor. Selected pressure-flow values were then numerically displayed along with the corresponding velopharyngeal orifice area calculation.

Procedure

The model was driven with varying airflow rates supplied by a compressed air source. Overall flow rates ranged from approximately 0.05 to 0.35 LPS and were selected to sample a wide range of flows known to exist during speech. Simultaneous pressure-flow measurements were obtained for each modeled velopharyngeal orifice area under each of the four conditions at airflow peak loci. These measurements were used to calculate the size of the velopharyngeal orifice opening using Warren's hydrokinetic equation (Warren and DuBois, 1964):

$$A = \frac{\dot{V}}{0.65\sqrt{2\frac{P_1 - P_3}{D}}}$$

where A is orifice area (cm^2), $\dot{V}$ is volume rate of airflow through the orifice, P_1 is pressure measured below the orifice, P_3 is pressure measured above the orifice, D is density of air, and 0.65 is a correction factor or constant term. In addition, measurements of velopharyngeal orifice area were used to compute percent error in calculated velopharyngeal orifice areas using the formula:

$$\text{Percent error} = \frac{KA - CA}{KA} \times 100$$

where KA is the known area and CA is the calculated area.

Twenty velopharyngeal orifice area and percent error calculations were obtained for the normal nasal resistance condition (condition #1), and 40 orifice area and percent error calculations were obtained for conditions of increased nasal airway resistance (conditions #2 to #4). The following three measures were obtained for each known orifice area under each of the four conditions: (1) mean orifice area, (2) standard deviation in calculated orifice area, and (3) percent error.

RESULTS AND DISCUSSION

The accuracy of velopharyngeal orifice area estimations obtained under conditions of normal nasal airway resistance (condition #1) and under conditions of elevated nasal airway resistance (conditions #2 to #4) are summarized in Tables 1 and 2. The data in Table 1 show that, in general, average calculated orifice areas corresponded favorably with orifice openings known to be present in the model. In addition, variation (standard deviation) in predicted orifice areas was small for all known orifice openings and conditions. Although the overall average predictive accuracy of velopharyngeal orifice area estimation decreased when nasal airway resistance was increased, the magnitudes of orifice estimation errors found in this study are similar to those established in previous modeling studies (Smith and Weinberg, 1980, 1982, 1983; Smith et al, 1984).

The data in Tables 1 and 2 show that a notable increase in percent error in prediction occurred for all conditions at the smallest orifice area (3.10 mm^2). An increase in error in prediction (overestimation) at small orifice openings has been reported previously (Smith and Weinberg, 1983; Smith et al, 1984). This overestimation is probably related to the observation that as orifice area decreases airflow speed increases and results in turbulent loss of energy. A turbulent dissipation term was neglected in the deri-

TABLE 1. Calculated Velopharyngeal Orifice Area Means (mm^2) and Standard Deviations for Known Orifice Openings Obtained Under Four Conditions of Increased Nasal Resistance

Known Orifice Area (mm^2)	*Condition #1*		*Condition #2*		*Condition #3*		*Condition #4*	
	$\bar{X}$	*S.D.*	$\bar{X}$	*S.D.*	$\bar{X}$	*S.D.*	$\bar{X}$	*S.D.*
3.10	3.59	0.08	3.69	0.08	3.70	0.08	3.76	0.07
6.96	7.25	0.35	7.50	0.26	7.37	0.25	7.56	0.13
12.55	12.57	0.41	12.19	0.26	12.60	0.36	13.30	0.34
19.72	19.43	0.44	20.21	0.54	18.94	0.25	18.52	0.34
28.39	27.44	0.47	29.56	0.92	27.10	0.46	26.40	0.82
38.12	40.54	1.44	39.53	1.32	41.89	1.19	30.94	1.01

TABLE 2. Mean Percent Errors in Prediction for Known Orifice Openings Under Increased Magnitudes of Nasal Resistance

Known Orifice Area (mm^2)	*Condition #1*	*Condition #2*	*Condition #3*	*Condition #4*
3.10	15.55	18.80	19.12	21.13
6.96	4.16	7.75	5.85	8.63
12.55	0.18	2.86	0.38	5.95
19.72	1.43	2.49	3.92	6.06
28.39	3.37	4.10	4.55	7.01
38.12	6.35	3.68	9.89	18.84
	$\bar{X}$ = 5.17	$\bar{X}$ = 6.61	$\bar{X}$ = 7.28	$\bar{X}$ = 11.27

vation of the hydrokinetic equation (see Appendix, Warren and DuBois, 1964). If such a term were included in the equation, the effective pressure differential would be greater and would lead to smaller orifice area calculations (i.e., less overestimation) and thereby more precise estimation. In the present study, the difference between the smallest known area (3.10 mm^2) and the calculated area associated with the largest overestimation (3.76 mm^2) is minute (0.66 mm^2) and is not likely to be meaningful in terms of clinical management or research investigation of velopharyngeal function.

A notable underestimation (30.94 mm^2) of the known orifice area (38.12 mm^2) occurred under the most resistive condition (#4). This may be explained by the observation that the plugs in the model nostrils in condition #4 created a very small cross-sectional area of the model nose and were located close to the nasal pressure-sensing catheter. Under these circumstances, spurious back pressures may have been created which resulted in less accurate orifice area calculations. This finding supports the observations of previous investigators (Warren et al, 1969; Smith et al, in press) that increased nasal obstruction may have deleterious effects on pressure-flow patterns for speech production and nasal respiration. Taken together, these findings suggest that nasal obstruction should be assessed and treated (if necessary) prior to assessment and physical management of velopharyngeal disturbance.

In summary, our data suggest that Warren's pressure-flow approach can provide accurate estimation of velopharyngeal orifice areas during speech in the presence of increased magnitudes of nasal airway resistance. These findings, together with those of previous studies (Smith and Weinberg, 1980, 1982, 1983; Smith et al, 1984; Smith et al, in press), support the view that pressure-flow assessment that incorporates hydrokinetic principles provides a useful noninvasive method for clinical testing and research investigation of velopharyngeal function. Future studies should include determination of the accuracy of velopharyngeal orifice area estimation in a more representative model of the human vocal tract with respect to vocal tract contours and velopharyngeal orifice length. Such a project is currently underway in our laboratory.

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