Effect of inferior surface angle on the self-oscillation of a computational vocal fold model

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Geometry of the human vocal folds strongly influences their oscillatory motion. While the effect of intraglottal geometry on phonation has been widely investigated, the study of the geometry of the inferior surface of the vocal folds has been limited. In this study the way in which the inferior vocal fold surface angle affects vocal fold vibration was explored using a two-dimensional, self-oscillating finite element vocal fold model. The geometry was parameterized to create models with five different inferior surface angles. Four of the five models exhibited self-sustained oscillations. Comparisons of model motion showed increased vertical displacement and decreased glottal width amplitude with decreasing inferior surface angle. In addition, glottal width and air flow rate waveforms changed as the inferior surface angle was varied. Structural, rather than aerodynamic, effects are shown to be the cause of the changes in model response as the inferior surface angle was varied. Supporting data including glottal pressure distribution, average intraglottal pressure, energy transfer, and flow separation point locations are discussed, and suggestions for future research are given.

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I. INTRODUCTION

In human vocal fold flow-induced oscillation during voiced speech, vocal fold geometry plays an important role in governing the tissue vibratory response as well as in influencing the pressure distributions that drive self-oscillation. Consequently, not only is geometry important in healthy phonation, but marked changes in voice quality have been associated with surgical procedures that have altered vocal fold geometry (e.g., Smith *et al.*, 1993). It is therefore important to understand the relationships between vocal fold geometry and vibration.

As illustrated in Fig. 1, the airway around the vocal folds can be decomposed into subglottic, glottic, and supraglottic regions. Prior research has shown that during vocal fold vibration, the glottal angle (ϕ , Fig. 1) varies as the mucosal wave propagates superiorly and as the glottal profile alternates between convergent and divergent orientations. The prephonatory glottal angle can also vary and influence the ease of phonation. For example, studies of flow through a glottis of different angles showed that some angles yield more favorable flow conditions than others for initiating and sustaining vocal fold vibration (Chan *et al.*, 1997; Lucero, 1998). Other studies have reported glottal pressure distributions in models with varying glottal angles, including symmetric and asymmetric glottal configurations (Scherer *et al.*, 2001; Shinwari *et al.*, 2003; Li *et al.*, 2006b). Some of the same studies have shown how convergent and divergent glottal angles relate to factors such as flow resistance, diffuser efficiency, and flow separation location (Hofmans *et al.*, 2003; Li *et al.*, 2006b).

As is the case with the glottal angle, the inferior surface profile (θ , Fig. 1) may also vary. Measurements made on laminagraphic tracings of the vocal fold contours taken during phonation have revealed that the inferior (as well as the superior) surface angle varies somewhat during the phonatory cycle, and that the average angle over the cycle varies significantly between human subjects (e.g., Agarwal *et al.*, 2003). Studies with excised canine larynges, distributed mass models, and finite element models also show motion of the inferior and superior vocal fold surfaces throughout the oscillatory cycle (Saito *et al.*, 1981; Baer, 1981; Titze, 1981; Fukuda *et al.*, 1983; Alipour *et al.*, 2000). Importantly, surgical procedures, particularly of the medialization type, can cause considerable changes to the subglottal profile at the inferior edge of the vocal folds (Grisel *et al.*, 2010).

While the aforementioned studies have shown that variations in inferior surface profile exist, few studies have been performed to quantify its influence on vocal fold vibration. Li *et al.* (2006a) examined how both the inferior and superior surface angles affected glottal pressure distributions in computational vocal fold models. In two other studies the effect of altering the subglottal profile on turbulence production in the glottis was explored using experimental models (Oren *et al.*, 2009; Grisel *et al.*, 2010). These three studies were all performed using only static models.

The purpose of this study was to explore how parametrically altering the inferior surface angle affected the flow-induced response of a computational self-oscillating vocal fold model. Output variables including model motion, flow rate, intraglottal pressure, and energy transfer were analyzed to determine the sensitivity of the model to changes in inferior surface angle. In the following sections the model and case studies are described, the results are presented and discussed, and suggestions for future work are provided.

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FIG. 1. Geometry of the human larynx from (top left) sagittal and (top right) coronal planes (adapted from *Gray's Anatomy of the Human Larynx*, www.bartleby.com, used with permission). (Bottom) Idealized vocal fold outline with convergent prephonatory glottal angle.

II. METHODS

A two-dimensional finite element model with distinct but fully coupled air flow (fluid) and vocal fold (solid) domains was created using the commercial fluid-structure interaction solver ADINA (ADINA R&D, Inc.) ADINA has previously been used in other studies of vocal fold flow-induced vibration (e.g., Thomson *et al.*, 2005; Decker and Thomson, 2007). The model was two-dimensional and utilized lateral symmetry for computational efficiency.

The solid domain geometry and material properties are shown in Fig. 2. It included four material layers, with an idealized exterior geometry based on the “M5” model of Scherer *et al.* (2001) with the modification of a $\phi = 4^\circ$ (ref. Fig. 1) convergent-shaped glottis. The four layers generally represented (1) the epithelium, (2) the superficial layer of the lamina propria (Reinke’s space), (3) the intermediate and deep layers of the lamina propria (ligament), and (4) the muscle (body) (Hirano *et al.*, 1981). To allow for large displacement and large strain, a hyperelastic Ogden material with nonlinear stress-strain properties was defined for the superficial lamina propria, ligament, and body layers. The shape of the stress-strain curves followed the equation

$$\sigma(\varepsilon) = A(e^{B\varepsilon} - 1), \quad (1)$$

used by Alipour-Haghighi and Titze (1991) to model the high strain portion of the stress-strain curve for vocal fold tissue. Here different values of A were used for the different layers while B remained constant. A value of $B = 10.5$ was found to reasonably approximate the shape of vocal fold lamina propria stress-strain curves reported by Chan *et al.* (2007). Values of $A = 22.5$, 112.7, and 839.4 Pa for the su-

FIG. 2. (Top) Solid and (bottom) fluid domains, including material properties, for computational model with 40° inferior surface angle. The epithelium thickness was $50 \mu\text{m}$. See text for material property details.

perficial lamina propria, ligament, and body layers were chosen, which corresponded to elastic moduli of 0.4, 2, and 14.9 kPa at 5% strain ($\varepsilon = 0.05$) for the three respective layers. The epithelium was defined as a linearly elastic material with a Young’s modulus of 50 kPa; however, it is acknowledged that this is only an estimate and that further research quantifying the range of modulus values that are characteristic of human vocal fold epithelium tissue is needed. Damping in all layers was estimated using the Rayleigh scheme with constants $\alpha = 56.549$ and $\beta = 3.979 \times 10^{-5}$, yielding a damping ratio around 0.05 in the frequency range between 100 and 300 Hz. The density and Poisson’s ratio values for all layers were 1070 kg/m^3 and 0.49, respectively. These material properties are comparable to those found in human vocal fold tissue and to those that have been defined in other similar simulations. As is shown below, in this study they were found to yield a model that exhibited a reasonable response in terms of glottal width amplitude (on the order of 1 mm), frequency (around 229 Hz), and vibratory motion (mucosal wave-like motion with an alternating convergent-divergent glottal profile).

The fluid domain (also shown in Fig. 2) consisted of subglottic, glottic, and supraglottic sections. Air with a density of 1.2 kg/m^3 and a viscosity of $1.8 \times 10^{-5} \text{ Pa s}$ was used. A pressure of 600 Pa was applied to the inlet. The outlet pressure was set to zero. A fluid-structure interaction boundary condition was applied at the fluid-solid interface along the vocal fold model wetted perimeter. With this boundary condition, the fluid and the solid domains were solved such that the displacement and stresses along the wetted boundary were equal (that is, the displacement and stress values along the fluid domain boundary were equal to the displacement

and stress values along the solid domain boundary) (ADINA, 2009). This boundary condition was applied with the no-slip condition, i.e., the velocity of the fluid along the boundary was the same as that of the solid. The symmetry line was defined using a slip-wall condition, and the remaining fluid domain lines were defined using no-slip wall conditions. The flow solver was based on the two-dimensional, unsteady, viscous, laminar, incompressible Navier-Stokes equations. The initial gap between the vocal fold medial surface and the symmetry line was 0.05 mm. To prevent total collapse of the fluid domain mesh during vibration, a rigid contact line was used to prevent the solid model from moving to within 0.02 mm of the symmetry line; this resulted in a minimum allowed total glottal width of 0.04 mm.

As shown in Fig. 3, the geometry was parameterized to create cases with five different inferior surface angles ranging from 15° to 45°. These were the same angles used by Li *et al.* (2006a) which were based on a range of values found in humans. Other geometric features shown in Fig. 2 (e.g., cover layer thickness, exit and entrance radius values) remained constant.

In order to consider both aerodynamic and coupled fluid-solid dynamic consequences of changing the inferior surface angle, three types of simulations were performed for each of the five angles. First, fluid-structure interaction (FSI) simulations were performed in which the complete vocal fold model was allowed to interact with the airflow. These are here referred to as the “fully flexible” cases. Second, simulations of just the airflow were performed using rigid vocal fold models, as was done by Li *et al.* (2006a) (“static” cases). The goal of these simulations was to isolate any effects on the aerodynamic flow that might be introduced only by changes in inferior surface angle, as well as to obtain static pressure distributions for comparison with pressure distributions from the fully flexible cases. Third, “partially rigid” cases were studied allowing for the influence of subglottal aerodynamics on model vibration to be investigated. This was done by separating the vocal fold model into deformable and rigid regions as shown in Fig. 4. The rationale for this third approach was as follows. Because the model geometry will change if the inferior angle is changed, the model mass and overall stiffness will also change. This will naturally cause a difference in vibration solely due to struc-

FIG. 3. Vocal fold surface profiles for five inferior surface angles.

FIG. 4. Partially rigid vocal fold model, showing (left) 25° and (right) 40° cases. Shaded area represents rigid portion.

tural considerations. In order to determine the degree to which changes in model vibration patterns could be attributed solely to changes in subglottal aerodynamics, in the third set of simulations, the upstream portion that included changes in inferior angle was rigid, whereas the geometry of the portion that was allowed to vibrate remained unchanged. The deformable region was the same as a vocal fold model with a 10° inferior surface angle. The solid region filled in the space of the remaining vocal fold geometry defined by a desired inferior angle. FSI simulations of the partially rigid cases were performed for each angle using the same parameters as the fully flexible simulations.

Simulations were performed to ensure that the results were reasonably independent of grid density, time step size, and convergence criteria. These were performed using the 45° fully flexible case. The study included sequentially varying the grid density by factors of two, the time step size by factors of 2, and the convergence criteria by factors of 10. Results were compared by analyzing glottal width waveforms, where glottal width was defined as twice the minimum lateral distance between the vocal fold medial surface and the symmetry line. Figure 5 includes results of the grid-independence study. Figure 5(a) shows glottal width vs time for three grid densities during the first 0.05 s. While the results were not entirely independent of grid density, the responses were very similar, as can be seen in Fig. 5(b), in which the waveforms for all three grid densities are shown over a normalized time period, T . Thus, in this case, the results of the model grid simulations and the other grid simulations were deemed to be satisfactorily close. Results for the time step size and convergence criteria independence studies are shown in Fig. 6. Varying these factors yielded considerably less change in glottal width waveforms than varying grid density. The final parameters were as follows. The fluid domain meshes (Fig. 7) consisted of approximately 32 000 nodes for all cases, maintaining the same mesh density for all cases. The solid domain meshes (Fig. 8) were of fairly uniform density throughout the domains, and again the same density was used in all cases. They contained approximately 42 000 to 54 000 nodes, depending on the inferior surface angle. Solutions for 6000 time steps with a time step size of 2.5×10^{-5} s were obtained.

III. RESULTS

A. Overview of model responses

For the fully flexible FSI cases, self-sustained oscillation was achieved for all but the 15° case. Instantaneous medial

FIG. 5. Grid independence study results, comparing minimum glottal width (a) for three grid densities during the first 0.05 s of the simulations and (b) waveforms over a normalized steady-state cycle. Model grid density (solid line), grid-halved density (short dashes), grid-doubled density (long dashes).

surface profiles are shown in Fig. 9 for six phases of the oscillatory cycle. Included for reference is the steady profile of the 15° case. The six phases shown in Fig. 9 correspond to instances of significance pertaining to motion and energy

FIG. 6. (a) Time step independence study results, comparing glottal width for three time step sizes. Model time step (solid line), doubled time step (short dashes), halved time step (long dashes); (b) convergence criteria independence study with three different convergence criteria showing results that are graphically indistinguishable.

FIG. 7. (Top) Fluid domain mesh, with view of entire domain and (bottom) close-up view of glottic region.

transfer that are discussed below. The first phase ($t/T=0$) corresponds to a “closed” glottis, or the time at which the glottal gap was at its minimum. The second and third phases ($t/T=0.2$ and 0.35) were during glottal opening. The maximum opening and the point of transition between opening and closing motion occurred at the fourth phase ($t/T=0.6$). The fifth and sixth phases ($t/T=0.73$ and 0.85) were during glottal closing.

Vibration frequencies were 225, 229, 228, and 235 Hz for the 45°, 40°, 35°, and 25° cases, respectively. Frequencies were obtained by inverting the average cycle period, which was calculated by averaging the times between ten successive peaks in the glottal width waveform. The vibration took place primarily in the medial portion of the cover with an inferior-superior propagating wave-like movement along the medial surface, creating a glottal profile that alternated between a convergent shape during opening and a divergent shape during closing. The 15° case yielded a divergent glottal profile and a steady-state superior displacement that was significantly greater than for the other cases.

The vibratory motion was similar between the 45°, 40°, and 35° cases (ref. Fig. 9). For these cases inferior-superior displacement was minimal and the alternating convergent-divergent glottal shape was apparent along the entire medial surface. The streamwise location of the minimum glottal width appeared to be essentially the same at all phases for these three cases. The only noticeable difference among these cases was a slight increase in superior displacement of the profile as the inferior angle decreased. Greater differences were seen in the 25° case vibration pattern: there was a more substantial increase in superior model displacement, a smaller portion of the medial surface (roughly the upstream half) was involved in the alternating convergent-divergent profile change, and the downstream half of the medial

FIG. 8. (Left) Solid domain mesh, with view of entire domain and (right) close-up of the medial portion of the vocal fold model.

FIG. 9. (Color online) Model surface profiles at six normalized times throughout the oscillatory cycle. Dashed lines show contact line position. Circular markers denote flow separation points.

surface remained in a basically divergent orientation throughout the cycle.

Comparisons are here made of the current model response with that of the human vocal folds and previous numerical and synthetic self-oscillating models. Data from the model with a 40° inferior surface angle was used for comparison, since this geometry has been commonly used in other studies. The output variables that are considered include glottal width, flow rate, frequency, pressure, and qualitative motion. These are discussed in further detail in the following sections, but are briefly used here for model comparison purposes.

Table I shows vibratory characteristics of the present model compared to data measured from the human vocal folds. In general, the present model exhibited features that compare well with actual human vocal fold vibration. Inlet pressure used here was in the range of typical subglottal pressures for softer phonation (e.g., Jiang and Titze, 1993). A maximum glottal width of 0.95 mm was very similar to *in vivo* measurements of about 1 mm (Schubert et al., 2002) and on the same order of magnitude as measurements of up to 4 mm from excised larynges (Doellinger and Berry, 2006; Boessenecker et al., 2007). The average flow rate of 178 ml/s fit within the range from 100 to 400 ml/s for average flow rate in human phonation (Jiang and Titze, 1993; Doellinger and Berry, 2006; Boessenecker et al., 2007). It is noted, however, that the flow rate for the present model was obtained by multiplying the two-dimensional flow rate data by 0.015 m (a representative length of the human vocal folds) in order to approximate three-dimensional data. This

is only an approximation since in humans, the glottis is not rectangular and the velocity profiles are not uniform in the anterior-posterior sense. Therefore, this value should only be interpreted as an estimate of what would be expected in a three-dimensional model. Vibration frequency was in the typical range of the human female speaking voice (e.g., Schubert et al., 2002). The alternating convergent-divergent glottal profile observed in the present model resembled the mucosal wave that travels along the medial surface of the human vocal folds during phonation (e.g., Boessenecker et al., 2007) and that has been shown to be an important factor in sustained self-oscillation of the vocal folds (e.g., Hirano, 1981; Titze, 1988).

Table II gives data for vibratory factors of some recent self-oscillating computational and synthetic models. A description of each model with the number of material layers, the type of geometry employed, and the numerical solver or synthetic material, is provided. Models with “M5 geometry” had geometry similar to the present model. The results of the present model agree reasonably well with and offer some improvements over these other models. Glottal width and flow rate of the present model fit in lower end of the range of glottal width and flow rate values for previous models. The fact that many of the previous models yielded higher values for these factors was possibly due to generally higher operating pressures for the previous models. Frequencies for the present model were higher than for previous models, possibly because of its nonlinear stress-strain properties. The well-defined alternating convergent-divergent glottal profiles due to

TABLE I. Data comparing the vibratory behavior of the model used in this study (present model) with human vocal fold vibration.

	Glottal Width (mm)	Flow Rate (ml/s)	Frequency (Hz)	Pressure (kPa)	Motion
Human Vocal Folds	1–4	100–400	120–240	0.4–3	Alternating convergent-divergent glottal profile due to mucosal wave
Present Model (40°)	0.95	178	229	0.6	Alternating convergent-divergent glottal profile due to mucosal wave-like motion

TABLE II. Data of vibratory factors from selected previous self-oscillating models for comparison to vibratory behavior of present model.

Model Description	Authors	Glottal Width (mm)	Flow Rate (ml/s)	Frequency (Hz)	Pressure (kPa)	Motion
Computational Models						
3-layer; continuum models coupled with flow predictions	Alipour <i>et al.</i> (2000); Alipour and Scherer (2000)	0.9–2.0	167–309	113–147	0.8 or 1.6	Convergent-divergent glottal profile due to wave-like motion
1-layer; 2D M5 geometry; ADINA FSI	Thomson <i>et al.</i> (2005)	4	–	93	2	Briefly convergent, mostly divergent glottal profile due to model deflection (no mucosal wave)
3-layer; 2D geometry from CT scan; immersed boundary method	Zheng <i>et al.</i> (2009); Luo <i>et al.</i> (2009)	0.9–1.5	266–322	160–230	0.8–1.2	Slightly convergent-divergent glottal profile
2-layer; 2D M5 geometry; ADINA FSI	Pickup (2010)	1.0–1.4	150–250	113–122	0.9	Briefly convergent, mostly divergent glottal profile due to model deflection (no mucosal wave)
Synthetic Models						
1-layer; M5 geometry; Silicone	Thomson <i>et al.</i> (2005)	3.5	169	120	2	Briefly convergent, mostly divergent glottal profile due to model deflection (no mucosal wave)
2-layer; M5 geometry; Silicone	Riede <i>et al.</i> (2008); Murray (2011)	1.5–4.0	550–750	100–120	0.8–1.0	Briefly convergent, mostly divergent glottal profile due to model deflection (no mucosal wave)
2-layer; MRI-based geometry; Silicone	Pickup and Thomson (2010); Murray (2011)	2.0–2.5	800–900	130–140	1.7–2.2	Alternating convergent-divergent glottal profile due to mucosal wave-like motion
4-layer; M5 geometry; Silicone	Murray (2011)	1.2–2.5	250–550	85–100	0.3–0.4	Alternating convergent-divergent glottal profile due to mucosal wave-like motion

mucosal wave-like motion of the present model were an improvement over the behavior of several previously used models and were similar to the motion of some more recent synthetic models.

B. Glottal width

Steady-state glottal width waveforms and glottal amplitude as a function of inferior surface angle are shown in Fig. 10. The horizontal axis of the waveform plot [Fig. 10(a)] is scaled by period (T) and phases have been adjusted such that the glottal width peaks are aligned. For reference, steady state glottal width for the 15° case was 0.47 mm (not shown). Glottal width was essentially the same for the 45° and 40° cases. These cases reached an amplitude of about 0.96 mm and full closure (i.e., touching the contact line for a minimum glottal width of 0.04 mm) for approximately 15% of the cycle. The 35° case amplitude was slightly smaller (about 0.88 mm) and was closed for a shorter portion of the period. The most significant difference is observed in the response of the 25° case, in which the amplitude was 0.67 mm (about 30% smaller than that of the other cases) and complete glottal closure was never achieved (the minimum glottal width was 0.16 mm). Figure 10(b) shows the changes in glottal amplitude with changing inferior surface angle. With angles less than 40°, glottal amplitude rapidly decreased as inferior angle decreased.

C. Flow rate

Flow rate waveforms, along with peak and average flow rates versus inferior surface angle, are shown in Fig. 11. In

the flow rate waveforms [Fig. 11(a)] the maximum flow rate occurs near, but slightly after, the time of maximum glottal width. Similar differences in flow rates were seen as with glottal widths. Peak flows were 440 and 437 ml/s for the 45° and 40° cases, respectively. A slight decrease (9%) occurred for the 35° inferior case and a greater decrease (30%) occurred for the 25° case. Figure 11(b) shows this trend. Flow rates at glottal closure were around 40 ml/s for all cases except 25°, which was 122 ml/s. Average flows over the cycle, also included in Fig. 11(b), were 198 ml/s for the 25° case and around 177 ml/s for the others.

D. Partially rigid model responses

Self-oscillation occurred for all inferior angle cases of the partially rigid model. The vibration frequency was close to 242 Hz for all cases, which is slightly higher than for the fully flexible cases. Vibratory motion also differed from that of the fully flexible cases. Motion was restricted in the inferior third of the medial surface because of the attached rigid portion, while the superior two-thirds alternated between a convergent and slightly divergent profile. In comparing the medial surface motion among the inferior angle cases for these models, no noticeable differences were observed. Figure 12 shows glottal width and flow rate for these simulations. Both glottal width and flow rate amplitudes were less than those seen for the fully flexible cases. The waveforms are essentially identical for all the cases, confirming that there are no differences in vibratory motion among the cases. The only deviation is a slight time shift between the 45° and the other cases; otherwise, the waveform shapes and frequencies were remarkably similar.

FIG. 10. (a) Steady-state glottal width for fully flexible model simulations. Vertical dotted lines correspond to phases shown in Fig. 9; (b) corresponding glottal amplitude vs inferior surface angle.

E. Energy transfer

For self-oscillation of the vocal folds to occur, there must be a sufficiently positive net transfer of aerodynamic energy from the airflow to the vocal folds to overcome damping within the vocal fold tissue (Titze, 1988). Energy transfer is thus directly related to vocal fold vibration. In order to further understand the physical mechanisms underlying how the inferior surface angle affects vocal fold vibration, aerodynamic energy transfer in the various cases is here discussed. Figure 13 shows steady-state aerodynamic energy transfer rates (power) and average intraglottal pressures for the fully flexible cases. Energy transfer rate is the product of force and velocity, and was here calculated as

$$\dot{E}_p = \int_s u_i (-p \delta_{ij}) (-n_j) dS, \quad (2)$$

where $\dot{E}_p$ is the energy transfer rate due to normal stress on the vocal folds with units of power (Joules per second), u is the fluid velocity (also equivalent to the vocal fold surface velocity at the fluid-solid interface owing to the no-slip condition), p is the fluid pressure, n is the control surface out-

FIG. 11. (a) Flow rate over steady-state cycle of fully flexible model simulations; (b) peak and average flow rate values vs inferior surface angle.

ward normal unit vector (points out of the fluid domain), and S represents the control surface (i.e., the vocal fold surface). The integrand, denoted by $\dot{I}_p$, is equivalent to intensity or energy flux (energy per unit time and per unit area). A positive energy flux ($\dot{I}_p$) occurs when the normal pressure acting on the vocal folds is in phase with the component of velocity normal to the vocal fold surface. Following the analysis of Thomson *et al.* (2005), energy transfer rate due to viscous stresses is here assumed to be negligible.

The general pattern of energy transfer rate to the vocal fold models included alternating positive and negative values during glottal opening, higher values during the closing portion of the cycle, and nearly no energy transfer at the times of maximum and minimum glottal width. Average intraglottal pressure can be seen to have correlated relatively well with power, i.e., varying pressures occurred within the first half of the cycle, and negative pressure occurred during closing at the times of peak power. Positive energy transfer during opening was produced when the general motion was lateral (a positive normal velocity component), and the average intraglottal pressure was positive. During closing the negative average pressure coupled with closing motion (negative normal velocity component) again resulted in positive

FIG. 12. Glottal width and flow rate waveforms for a steady-state cycle of partially rigid model simulations.

energy transfer. At the points of maximum and minimum glottal width, the major contributor to low energy transfer was the low surface velocity of the vocal fold associated with changing direction of the surface motion.

As with the other output measures discussed above, while little difference in energy transfer rate behavior was observed among the 45°, 40°, and 35° cases, greater dissimilarity existed between these and the 25° case. For the three highest angles, the waveforms and peak power values were comparable, with the 40° case showing a slightly higher peak than for the other cases and the 35° case varying only slightly from the other cases throughout the cycle. Compared to these cases, the major difference apparent in the 25° case was lower amplitude. While the waveform bore a similar shape to the higher angle cases, amplitudes of both positive and negative energy transfer rate were generally lower throughout the cycle. Total cycle energy transfer values (integrals of energy transfer rate over one period, shown in Fig. 14) also followed this trend. The greatest energy transfer of 1.22×10^{-5} J was seen for the 40° case. The 45° and 35° cases yielded comparable values (1.21×10^{-5} and 1.00×10^{-5} J, respectively), while the total energy transfer was considerably less for the 25° case (0.23×10^{-5} J).

FIG. 13. (Top) Aerodynamic energy transfer rates (power) and (bottom) average intraglottal pressures for fully flexible cases over a steady-state cycle. Vertical dotted lines correspond to phases shown in Fig. 9.

Inspection of vocal fold surface velocity, pressure distribution, and energy flux ($\dot{I}_p$) at three particular phases (Figs. 15–17) provides further insight into the model responses. Figure 15 includes plots of these data for the 40° and 25° cases at the $t/T=0.2$ opening phase (plots of these cases at $t/T=0$ and $t/T=0.35$, not shown, were found to exhibit similar trends). As seen in Fig. 13, at this phase there was an energy transfer rate local maximum for both cases. Corresponding rigid model profiles and static model pressure

FIG. 14. Total energy transfer over one cycle for each self-oscillating inferior surface angle case.

FIG. 15. (Top) Medial surface position, (middle) pressure distribution, and (bottom) energy flux along the vocal fold model surface at $t/T=0.2$ during a steady-state cycle for 40° and 25° inferior angles. In the top row the gray lines along the medial surface denote surface velocity. The dashed curves in the top and middle rows denote static model position and pressure distribution, respectively.

distributions are included for comparison (the pressure distribution was nearly identical for all of the static cases). For the fully flexible cases, pressure distributions differed from the static distributions and were distinct between cases, contributing to differences in energy flux. Pressures for the 40° and 25° cases reached their respective minima at different locations. This corresponds to the difference in location of the point of minimum glottal width along the medial surface between the models.

For the 40° case most of the energy flux was produced in the glottis immediately upstream of the location of minimum width, where the surface velocity into the fold was in phase with the positive pressure. At the location of minimum width, pressure became slightly negative while velocity remained into the fold, resulting in the small negative energy flux just before the zero distance mark. On the inferior surface pressure was at essentially the inlet value, while the cover velocity alternated movement on different portions to create both positive and negative energy flux. For the 25° case at this phase, the behavior was similar to the 40° case, but as the convergent portion of the glottis was shorter and moved with a lower surface velocity, the less positive energy flux resulted.

The phase $t/T = 0.6$ (Fig. 16) was the point of maximum glottal opening and a point of almost no energy transfer for all of the cases. Here the pressure distributions were different from the static simulations in that there were two pressure local minima for both the 40° and 25° cases. These were associated with the glottis shape; note the dual elevated regions of the surfaces at both the entrance and exit for the 40° case, and a slight elevated region at the entrance with a larger elevated region in the center of the medial surface for the 25° case. Along the surface, energy alternated between slightly negative and positive flux values as pressure and velocity alternated between being in and out of phase. Integrating these alternating fluxes across the entire surface resulted in a net energy transfer rate that was close to zero.

Corresponding data at $t/T = 0.73$, the point of highest energy transfer for all the cases, are shown in Fig. 17. The pressure distribution for the 40° case included a highly negative pressure drop and a recovery pressure along the medial surface that was significantly less than zero. Notably, this pressure drop was much lower than the corresponding static pressure drop. This highly negative pressure coupled with high closing velocities created a positive energy flux that

FIG. 16. (Top) Medial surface position, (middle) pressure distribution, and (bottom) energy flux along the vocal fold model surface at $t/T=0.6$ during a steady-state cycle for 40° and 25° inferior angles. In the top row the gray lines along the medial surface denote surface velocity. The dashed curves in the top and middle rows denote static model position and pressure distribution, respectively.

was higher than at other phases and occurred over almost the entire vocal fold surface. Considerable positive energy flux also occurred on much of the inferior surface, further contributing to the energy transfer rate at this phase. The pressure distribution for the 25° case was also considerably negative, yet not nearly as great in magnitude as for the 40° case. As with the 40° case, the higher closing velocity was in phase with negative pressure, creating a significant amount of positive energy transfer. However, because less of the medial surface was involved in the movement for this case than for the higher angles and because pressures and velocities were not as great in magnitude, the peak positive energy transfer value was roughly five times lower than for the other cases.

F. Flow separation

The point of flow separation along the vocal fold surface is closely tied to surface pressure distributions (Alipour and Scherer, 2004; Thomson *et al.*, 2005) and hence the driving forces that govern vocal fold self-oscillation. For the present study it was of interest to investigate how flow separation

location was affected by changes in inferior surface angle. Separation locations at different phases were manually identified using flow visualization tools in the commercial post-processing software EnSight (CEI, Inc.) The circular markers in Fig. 9 show the estimated separation point locations. The points were closely grouped at all phases for the 45° , 40° , and 35° cases. During the first two phases, separation point for the 25° case was also grouped with the others. It is important to note that while separation points appeared close to the same location, because the models tended to deflect more superiorly and the point of minimum glottal width moved gradually upstream with decreasing angle, flow separation actually moved slightly upstream in relation to the position of the superior surface for each model as inferior angle decreased. At $t/T=0.35$ separation for the 25° case was significantly upstream of the other cases and then shifted downstream during closing. An interesting phenomenon was seen at $t/T=0.6$, where two separation points occurred for the 45° , 40° , and 35° cases. The two elevated surface regions at the glottis entrance and exit caused two distinct pressure drops (see Fig. 15), which caused flow separation on the recovery of each drop. While there were two slightly elevated

FIG. 17. (Top) Medial surface position, (middle) pressure distribution, and (bottom) energy flux along the vocal fold model surface at $t/T=0.73$ during a steady-state cycle for 40° and 25° inferior angles. In the top row the gray lines along the medial surface denote surface velocity. The dashed curves in the top and middle rows denote static model position and pressure distribution, respectively.

surface regions on the medial surface for 25° at $t/T=0.6$, the flow remained attached after the first peak, and there was only one separation point, located downstream of the second peak.

For comparison with previously reported data (Alipour and Scherer, 2004; Li *et al.*, 2006b; Decker and Thomson, 2007), separation point location is here related to two measures, one of which is related to pressure distribution, and the other is related to glottal width. For the former measure (here denoted p^*), the pressure drop at separation (Δp_{sep}) is expressed as a percentage of the total pressure drop (Δp_{tot}), i.e.,

$$p^* = \Delta p_{sep} / \Delta p_{tot} \times 100\%, \quad (3)$$

where Δp_{sep} is calculated as the difference between the downstream recovery pressure and the pressure at separation ($\Delta p_{sep} = p_{rec} - p_{sep}$), and Δp_{tot} is the difference between the recovery pressure and the minimum pressure ($\Delta p_{tot} = p_{rec} - p_{min}$). The recovery pressure, p_{rec} , is the highest pressure along the vocal fold surface after separation. Table III contains p^* values for all cases and phases. Average values of p^* over the six phases for the 45° , 40° , 35° , and

25° cases were 24.9%, 29.0%, 37.3%, and 38.1%, respectively. For the 15° case p^* was 45.7%.

For the other separation point measure related to glottal width, the area ratio

$$A^* = A_{sep} / A_{min} \quad (4)$$

is used, where A_{sep} is the glottal area at separation and A_{min} is the minimum glottal area (see, for example, Alipour and Scherer, 2004; Decker and Thomson, 2007). A^* values for all inferior angles and phases are listed in Table III. Most A^* values ranged from approximately 1.10 to 1.40. Values were considerably higher at $t/T=0$ (3.58, 3.73, and 3.48) and $t/T=0.86$ (3.51, 2.01, and 1.44) for the three highest angles. Average A^* values over all six phases were 1.90, 1.70, 1.58, and 1.25 for the 45° , 40° , 35° , and 25° cases, respectively, with $A^* = 1.13$ for the 15° case, showing a trend of decreasing A^* with decreasing angle.

These separation point measurements support the findings of previous studies. p^* values showed that separation points occurred upstream of total pressure recovery. This observation is consistent with general fluid dynamic theory as well as other voice production studies (Hofmans, 1998;

TABLE III. Flow separation point locations in terms of p^* and A^* for each case at six phases.

	$t/T=0$	$t/T=0.2$	$t/T=0.35$	$t/T=0.6$	$t/T=0.73$	$t/T=0.85$
	p^*					
45°	20.44%	37.07%	38.67%	7.14%, 18.38%	39.14%	13.71%
40°	20.98%	40.65%	32.80%	10.50%, 29.51%	38.55%	29.69%
35°	20.18%	42.41%	29.29%	31.26%, 34.94%	61.45%	41.73%
25°	39.31%	34.65%	43.49%	43.18%	31.32%	36.48%
15°	45.74%	45.74%	45.74%	45.74%	45.74%	45.74%
	A^*					
45°	3.58	1.39	1.16	1.21, 1.31	1.17	3.51
40°	3.73	1.40	1.20	1.20, 1.27	1.08	2.01
35°	3.48	1.36	1.24	1.15, 1.24	1.13	1.44
25°	1.35	1.27	1.17	1.14	1.17	1.41
15°	1.13	1.13	1.13	1.13	1.13	1.13

Shinwari *et al.*, 2003; Li *et al.*, 2006b). For example, Li *et al.* (2006b) reported values of $p^* = 6.5\%$, 13.0% , 41.9% , and 34.4% for static divergent models with intraglottal divergence angles of 5° , 10° , 20° , and 40° , respectively; these values are similar to the p^* values from the present self-oscillating models. A^* values presented here are similar to values in the range of 1.00 to 1.73 reported by Decker and Thomson (2007) and to the values in the range of about 1.1 to 1.9 with an average of 1.47 predicted by Alipour and Scherer (2004). The much higher A^* values seen in the present models typically occurred when the glottis was at its maximum closure, where the very small glottal width (A_{min}) caused a large increase in area ratio.

IV. DISCUSSION

The results showed that changing the inferior angle significantly influenced model behavior. The model response became less characteristic of human vocal fold vibration at the smallest inferior angles. While the changes were less noticeable from 45° to 35° , the 25° case showed that decreasing the inferior angle increased superior model displacement. With a smaller inferior angle, the overall mass and effective stiffness of the model was reduced. Given that each case was subjected to the same inlet pressure, it was natural that the models with less mass and stiffness would be deflected more in response to the flow. The increased deflection changed the overall glottal profile to a more divergent shape and also increased the pre-steady-state glottal width, factors which are known to adversely affect vibration (Lucero, 1998). Significant decreases in glottal width and flow rate amplitudes associated with decreases in inferior angle were also observed.

Changing the inferior surface angle clearly altered the structural response of the models by changing both mass and stiffness. The purpose of the partially rigid model simulations was to remove these effects and explore how vibration was affected solely due to upstream aerodynamic changes caused by changing the inferior angle. As was shown, there were essentially no differences in vibration among the cases for these models. This indicates that the inferior angle itself did not introduce any significant aerodynamic changes that altered vibration. This is consistent with the findings of Li

et al. (2006a) that pressure distributions were not affected with changing inferior angle, but the present study extends this conclusion to the self-oscillating case. It can therefore be concluded that the observed changes in vibration seen in the fully flexible cases were primarily due to changes in structural, not subglottal aerodynamic, responses.

In order for self-sustained vocal fold oscillations to occur, a positive net airflow-to-tissue energy transfer is required to overcome energy dissipation within vocal fold tissue (Titze, 1988; Thomson *et al.*, 2005). The present results showed a positive net energy transfer to the vocal folds for all self-oscillating cases. Generally, the energy transfer decreased with decreasing inferior surface angle. This was consistent with corresponding reductions in glottal width waveform amplitude.

Separation point location also plays a role in energy transfer to the vocal folds by affecting pressure distributions. If separation occurs more upstream than downstream, average pressures are generally lower and energy transfer is generally less. The general trend observed in these models was that the separation point moved upstream on the medial surface with decreasing angle, especially during glottal opening. This was consistent with the lower average intraglottal pressures during opening and less energy transfer for smaller angles.

Some possible limitations of the flow solver used in this study should be mentioned. For one, the flow model did not include turbulence. Other possible approaches include the use of different turbulence models, including solvers using Reynolds-averaged Navier-Stokes (RANS) models or large eddy simulation (LES). In general, different turbulence models can yield different predictions in terms of glottal jet dynamics and intraglottal pressure distributions (e.g., Mylavaram *et al.*, 2009). Additionally, in general, numerical models exhibit different degrees of numerical dissipation, a consequence of which is differences in predicted flow-related losses and fine-scale flow feature details. For the cases discussed here, it is likely that the intraglottal pressure distributions leading up to the point of flow separation predicted by various flow solvers and turbulence models would not deviate significantly. This is based on an expectation of a relatively undisturbed flow up to this point. Downstream of

flow separation, however, the flow field becomes more complex, the role of flow features on intraglottal pressure distributions may become more pronounced, and the choice of numerical solver may become more consequential. For example, LES modeling of glottal flow has predicted intraglottal vortices in the superior region of a divergent glottis, downstream of the separation point, with associated significantly negative local pressure minima on the vocal fold surface past the point of flow separation (Mihaescu *et al.*, 2010). In terms of the model described in this paper, it is not expected that the basic conclusions regarding model response (e.g., that glottal amplitude decreases with decreasing inferior surface angle) would vary significantly if another type of model were used, but that some of the details (e.g., flow rate wave form) may vary. In general, the sensitivity of the responses of self-oscillating computational vocal fold models to flow modeling choices is a topic in need of future exploration.

V. CONCLUSIONS AND FUTURE WORK

A two-dimensional, self-oscillating, finite element model of the vocal folds was used to study the effect of changing inferior vocal fold surface geometry on vocal fold vibration. Fully coupled fluid-structure interaction simulations with varying geometry were performed. Simulations using a partially rigid model were carried out in order to isolate and study the effects of changing subglottal aerodynamic loading on the model. Results of the fully flexible model simulations showed that model vibration was significantly influenced by variations in inferior surface angle. Specifically, reducing the angle resulted in greater vertical displacement and a smaller portion of the medial surface that participated in alternating convergent-divergent motion. As the inferior surface angle was reduced, the glottal width amplitude, flow rate amplitude, and net airflow-to-tissue energy transfer values all decreased. The partially rigid model responses showed that changes in vibration of the fully flexible models occurred primarily because of changes in structural, not aerodynamic, model responses.

In order to confirm these findings, draw further conclusions regarding vocal fold vibration, and explore the influence of changing inferior surface angle on voice quality, future studies are recommended. These include more extended computational modeling (e.g., three-dimensional models, turbulent airflow, more anatomically accurate geometry, and materially anisotropic models over a range of geometries) and experiments using synthetic vocal fold models and possibly excised larynges. Measurements quantifying human epithelium modulus would enable more accurate modeling, and studies that explore the sensitivity of model response to epithelium modulus could be insightful. Future research could also include in-depth exploration of the structural dynamic changes associated with inferior angle changes and investigation of optimal profile geometry.

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