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The effect of false vocal folds on laryngeal flow resistance in a tubular three-dimensional computational laryngeal model

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Abstract

Objective—The current study used a three-dimensional computational laryngeal model to investigate the effect of false vocal folds on laryngeal flow resistance.

Method—A three-dimensional, tubular shaped computational laryngeal model was designed with a high level of realism with respect to the human laryngeal anatomy. Two cases, one with the false vocal folds and the other without the false vocal folds, were created in the numerical simulation to compare the laryngeal flow behaviors.

Results and conclusion—The results were discussed in a comparative manner with the previous two-dimensional computational model. On one hand, the results demonstrated the similar mechanism as observed in the two-dimensional model that the presence of the false vocal folds suppressed the deflection of the glottal jet and in doing so, reduced the **mixing-related** minor loss in the supraglottal region. On the other hand, the three-dimensional flow was more stable and straighter, so the effect of false vocal folds on suppressing the jet deflection in the three-dimensional model was not as prominent as in the two-dimensional model. Furthermore, the presence of the false vocal folds also increased the friction-related major loss due to the increased velocity gradient in the restricted flow channel. Therefore it was hypothesized that the final effect of the false vocal folds on flow resistance is the combined effect of the reduced **mixing-related** minor loss and increased friction-related major loss, both of which are highly related to the gap between the false vocal folds.

Keywords

False Vocal Folds; Computational Laryngeal Model; Laryngeal Flow Resistance; Glottal Jet Deflection; Three-dimensional Glottal Flow Dynamics

I. Introduction

Laryngeal flow resistance is an important aerodynamic parameter regarding the efficiency of larynx. It is generally defined as $Z_L = \Delta P_L / \bar{Q}$, where ΔP_L is the mean pressure drop across

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the glottis and $\bar{Q}$ is the associated mean flow rate through the glottis. The effect of **false vocal folds (FVFs)** on laryngeal flow resistance has been examined in a number of studies with various computational and experimental approaches¹⁻⁴. For example, through direct numerical simulation of the compressible flow in a two-dimensional (2D), static, axisymmetric laryngeal model, Zhang et al.¹ found that the presence of the FVFs reduced the flow resistance. A similar conclusion was also drawn in our recent study using a 2D flow-structure interaction computational laryngeal model⁴. In that study we also proposed and proved that the dominant mechanism for the reduction of the flow resistance was the reduction of the minor loss associated with the flow mixing due to the suppression of the large scale jet deflection by the FVFs. This proposition was supported by another computational study which found that decreasing the FVFs gap resulted in higher flow resistance⁵. However, the aforementioned studies were all limited to 2D computational models. A recent computational study comparing the jet behavior through 2D and three-dimensional (3D) glottal orifices demonstrated that the 3D flow was more stable and consequently showed much weaker deflection compared to the 2D flow⁶. This finding implies that, in the 3D model, the effect of FVFs on suppressing the jet deflection may not be as prominent as in the 2D model. Therefore, their effect on reducing the flow resistance may not be as strong as in the 2D model.

On the other hand, Agarwal et al.² used a Plexiglas model of the larynx to experimentally examine the effect of FVFs on flow resistance. Their results suggested that the effect of FVFs on flow resistance depended on the ratio of FVFs gap to **true vocal folds (TVFs)** gap. For a ratio lower than one, the flow resistance was increased by the presence of the FVFs; for a very large ratio higher than 10, flow resistance was not affected by the presence of FVFs and for an intermediate ratio between 1 and 10, the flow resistance was reduced by the presence of the FVFs. For a ratio of about 2.0, the flow resistance was found to reach a minimum which was 20-25% lower than the resistance in the absence of the FVFs. Discrepancy between their results and our previous 2D computational results⁴ was noticed. The ratio of the FVFs to TVFs gaps in our 2D computational model was 12 and the flow resistance was reduced by 17% with the addition of the FVFs in that study, while, according to the results of Agarwal et al.², such high gap ratio (higher than 10) should not have effect on the flow resistance. It needs to be pointed out that the distance between the TVFs and FVFs was around 0.6cm in both models. The potential reason for this discrepancy could be the three-dimensionality of the flow, flow pulsatility or flow-structure interaction, and that has not been addressed yet. In a more recent study, Alipour et al.³ used an excised canine larynx to study the effect of FVFs. They based their conclusions on the differential flow resistance which was defined as $Z'_L = \partial(\Delta P_L) / \partial Q$. They concluded that adding the FVFs resulted in an increase in the differential translaryngeal flow resistance. In fact, if the same definition of the flow resistance as in Zheng et al.⁴ and Agarwal et al.² was used

$\left(Z_L = \Delta P_L / \bar{Q}\right)$, the flow resistance for many cases in Alipour et al.³ was still increased with the presence of the FVFs (see Table 2 and Figure 3 and 5 of their paper). Therefore, the experimental observations in Alipour et al.³ also contradicted the observations in 2D computational models.

The discrepancies between the experimental studies and 2D computational studies motivated the current study to use a 3D flow-structure interaction (FSI) computational laryngeal model to examine the effect of FVFs on laryngeal flow resistance. A 3D, tubular shaped laryngeal model that has been designed with a high level of realism with respect to the human laryngeal anatomy was utilized. As an extension of the previous 2D computational model⁴, the results were extensively compared to the 2D model.

II. Simulation Setup

The numerical algorithm and simulation set up have been reported in details in our previous studies⁷⁻⁸. For the sake of completeness, the current paper describes concisely some salient features of the numerical methods, geometric model, contact model, boundary conditions, and material properties.

The current study employed an explicitly coupled immersed-boundary-finite-element method based flow-structure interaction solver (FSI) to model human phonation⁸. The glottal airflow was governed by the 3D, unsteady, viscous, incompressible Navier-Stokes equations, and the TVF dynamics were governed by the Navier equations. The coupling between the flow and solid solvers was implemented by tracking the aerodynamic loading on the interface mesh as well as its deformed shape and velocity in a Lagrangian fashion.

The geometric model with the FVFs (denoted as (+)FVF case) is shown in Figure 1(a). The model consisted of a 12 cm long elliptical cylinder, a pair of deformable TVFs and a pair of rigid FVFs. The entire model was immersed in a 1.6cm×12cm×2.2cm rectangular computational domain. The cross-section profile of the TVF was generated based on a high resolution laryngeal CT scan of a p male subject⁷. A 0.01cm minimum glottal gap between the TVFs was enforced during the simulations to ensure the success of the flow solver. The overall shape and dimensions of the FVF were based on in-vivo measurements⁹ and shown in Figure 1(d). Figure 1 (c) is the superior view of the geometrical model of the vocal tract. The pair of FVFs was placed in to the elliptical vocal tract with an anterior angle of 23.5°, which was measured from the laryngealoscopic superior-to-inferior view of the larynx of a human subject⁸. The two FVFs were touching at the anterior end, and the gap between them gradually increased toward the posterior end. At the posterior end, the gap between the pair FVFs was around 0.65 cm. The FVFs was located from y=3.39cm to y=5.3 cm. The minimal gap between the FVFs was located at y=4.0 cm. The distance from the glottal exit to the minimal gap between the FVFs was 1.0 cm.

The configuration of the three inner layers of the TVF is shown in Figure 1(e), and it was constructed roughly based on the anatomical data¹⁰. The thickness of each layer was assumed to be constant along the anterior-posterior direction. The material of the three layers was assumed to be linear, viscoelastic, and transversally isotropic. The material properties for each layer are reported in Table 1. It should be noted that the TVFs barely vibrate in the longitudinal direction during phonation, and therefore we implemented an in-plane motion constraint by setting the longitudinal Young's Modulus in each layer equal to 10^{-4} time transversal Young's Modulus¹¹.

0.0 kPa and 1.0 kPa gauge pressure were applied at the exit and inlet of the tube, respectively. Non-slip and non-penetration boundary conditions were applied on all the walls. A zero displacement boundary condition was imposed on the lateral surfaces and a traction boundary condition was applied on the medial surface of the TVFs. The flow solver employed a high resolution, non-uniform $64 \times 256 \times 128$ Cartesian grid for the computational domain. The minimum glottal gap width was resolved by at least two points in the Z direction. The solid-dynamics solver employed 63117 tetrahedral elements to discretize each TVF.

The geometric model without the FVFs (denoted as (-)FVF case) was generated by removing the FVFs inside the vocal tract. The boundary conditions and grid resolution were kept same as the (+)FVF case. A small time-step corresponding to 3.5×10^{-6} s was employed for both simulations, and this resulted in about 1200 time steps per vibration cycle. The simulations were conducted for about 30 cycles to ensure that initial transients were eliminated from the analysis, and each simulation took about 1200 hours on 128 processors of an Intel 3.0 GHz Quad-Core IBM iDataplex parallel computer.

III. Results and discussion

Figure 2 shows the temporal variation of the glottal volume flow rate of the two cases. Some key flow quantities as well as their differences between the two cases are reported in Table 2. To further compare the differences between the 2D and 3D models, corresponding quantities in the previous 2D model⁴ are also provided in Table 2. As mentioned earlier, a minimum glottal gap was enforced in lieu of a complete glottal closure and this gap produced a small amount of leakage flow during glottal closure. To remove the effect of this artificial phenomenon in the comparison of the flow quantities, the flow rate was assumed to be zero when the glottal gap reached the minimum value (0.01cm).

It was found that the fundamental frequency (F_0) only increased 2.53% from (-)FVF to (+)FVF, suggesting that the presence of the FVFs did not have a noticeable effect on the fundamental frequency. This finding is in line with the 2D model which showed 2.2% increase in the fundamental frequency with the addition of the FVFs⁴. To better understand the effect of FVFs on the frequency components, the frequency spectra of the glottal flow rate over the sustained vibration stage were **computed by Fourier frequency transfer** for the two cases and are shown in Figure 3. It shows that the fundamental frequency and the first two superharmonics were same for the two cases. The energy density associated with these frequency components was also nearly equal for the two cases. However, after the second superharmonics, the spectrum of (-)FVF did not show clear superharmonics and the energy started to smoothly decay with frequency while the spectrum of the (+)FVF case still clearly showed the third and fourth overtones. It was also noticed that the energy associated with the nonharmonic frequencies as well as higher frequencies was significantly lower in the (+)FVF case than the (-)FVF case. Signal-to-noise ratio (SNR) was calculated for the two cases in Matlab and is reported in Table 2. SNR was defined as the ratio of the signal power to the noise power. It increased from 19.1 in the (-)FVF case to 35.6 in the (+)FVF case. These observations suggest that while the addition of the FVFs may not have

noticeable effect on the fundamental frequency and flow rate, it could increase SNR and therefore improve voiced quality.

The averaged peak volume flow rate (Q_{peak}) and time-mean flow rate $\bar{Q}$ are reported in Table 2. They both showed a slight increase from (–)FVF to (+)FVF, which were 2.75% and 3.14%, respectively. It was reported that the typical mean glottal flow rate measured in the experiments ranged from 110 to 220 ml/s¹² and the peak glottal flow rate ranged from 200 to 580 ml/s¹³. In the current study, the mean glottal flow rate was around 195 ml/s and the peak flow rate was around 300 ml/s. They were well within these typical ranges. By defining the translaryngeal flow resistance as $\Delta P / \bar{Q}$, the current model implied a 3.05% reduction in the translaryngeal flow resistance by adding the FVFs. This value was significantly lower than that in the 2D model which was 17% reduction in the flow resistance⁴.

In the 2D model, it was demonstrated that the primary mechanism for the reduction of flow resistance was the suppression of the large scale jet deflection due to the presence of the FVFs. The straightening of the jet reduced the flow mixing in the supraglottal region and hence reduced the viscous loss. It was of interest to examine if such mechanism was still observed in the 3D model. The phase averaged flow fields were computed for 20 equal-distant phases over the sustained vibration cycles. Figure 4 shows the contour of the velocity magnitude at the center plane of the larynx at the 1st, 3rd, 5th, 7th, 9th, 11th and 13th phases. These phases are also indicated in the glottal flow waveform to illustrate different development stages of the glottal jet. It was found that the glottal jet started to skew from the beginning of the cycle (phase 1 and 3) for both cases. This observation was in line with the 2D model and was demonstrated to be due to the entrainment of the remnant vortices near the glottal exit created in the previous cycle^{4,8}. Other than the deflection direction that the jet in (–)FVF deflected to the right and the jet in (+)FVF deflected to the left, the two jets showed similar behaviors: they experienced a spreading due to the transverse diffusion of the shear layer and the entrainment between the high momentum jet core and the surrounding flow; the deflection was accentuated as the jet developed. However, shortly afterwards (phase 5, 7 and 9), the two cases started to exhibit noticeable differences; the jet in (–)FVF continued to deflect toward the left side while the jet in (+)FVF attached to the right FVF and maintained relatively straight trajectory as it passed through the FVFs. The large deflection of the jet in (–)FVF also resulted in larger area of high velocity in the left side of the channel. It indicated that the existence of the FVFs suppressed the large deflection of the jet which was in line with the observation in the 2D model. At later stages of the cycle (phase 11 and 13) the flow quickly decelerated. It became unstable and more susceptible to the downstream disturbance. Therefore, deflection of the upstream portion of the jet was observed at this stage for both cases.

As demonstrated in the 2D model, the suppression of the large scale jet deflection reduced the mixing in the supraglottal flow and in doing so, reduced the viscous loss. In order to examine if such mechanism still existed in the 3D model, the phase-averaged turbulence kinetic energy (TKE) was calculated to measure the mixing level. Figure 5 shows the contour of the TKE at the mid-coronal plane for the two cases. Significant differences in TKE between the two cases can be observed at phases 7, 9 and 11 when the flow rate was

high. In the (+)FVF case, the peak value of TKE and the area of high TKE region were both significantly smaller than those in the (–)FVF case. Take phase 11 as an example. At this phase, the flow was quickly decelerating and had the highest mixing level throughout the cycle. The peak value of TKE was 0.13 in the (–)FVF case and 0.09 in the (+)FVF case, representing a 31% reduction with the addition of FVFs. The reduced mixing would reduce the associated viscous loss and so the flow resistance. Therefore, the flow resistance was lower in the (+)FVF case, as shown in Table 2. The reduced mixing also explained the reduced noise in the (+)FVF case. It may be worthwhile to mention that in classical singing the larynx is generally lowered, and so the distance for the flow to travel from the glottis exit to FVFs becomes longer. In this case, the effect of FVFs on reducing the flow deflection and flow mixing might become weaker. Therefore, it might result in more noises.

However, it was also noticed that the reduction of the flow resistance in the 3D model was not as significant as in the 2D model. It is important to explore the mechanism for this discrepancy between the 2D and 3D models. In our opinion, there might be three main mechanisms weakening the effect of FVFs in the 3D model. First, compared to the 2D jet, the 3D jet is more stable and less susceptible to the disturbance of downstream vortical structures; therefore the deflection of the jet will not be as strong as in the 2D model. It can be confirmed in Figure 4 that even in the (–)FVF case, the upstream part of the jet was mostly straight and only the downstream head showed large deflection. In contrast, in the 2D model, the deflection of the jet was much stronger. The jet was able to impinge the bound wall and further interact with the wall boundary layer to form a large rotating vortex in the supraglottal region. This rotating vortex significantly enhanced the mixing in the supraglottal area. Therefore, the role of FVFs in suppressing flow mixing was more prominent in the 2D model. Mattheus and Brucker⁶ had a comparison of glottal jet deflection between the 2D and 3D models and also found that the flow in the 2D model had much stronger deflection. Other experimental and computational studies also revealed a more stable 3D jet^{14–15}.

The second possible reason for the less prominent effect of the FVFs in the 3D model was that the 3D jet showed preferential deflection direction which can persist for many cycles while the 2D jet showed cycle-to-cycle variation of the deflection direction. Therefore, the high level of mixing was only localized in one half of the channel in the 3D model while it was in the entire channel in the 2D model. Thus, the effect of FVFs on suppressing mixing was expected to be more prominent in the 2D model.

Thirdly, it should be noticed that when the FVFs reduced the **mixing-related** minor loss, they, on the other side, increased the friction-related major loss due to the increased velocity gradient in the restricted area of the channel. The friction in the flow is caused by the viscosity when one layer of the fluid is moving with respect to an adjacent layer or against the wall of the channel. The amount of friction loss (or major loss) depends on the fluid viscosity and the velocity gradient of the flow. In the larynx, the FVFs restrict the flow channel in the supraglottal region, and therefore would increase the velocity gradient in that region. Figure 6 shows the time-averaged streamwise velocity across the channel at two locations at Y=3.2 cm and Y=4.0 cm (shown in Figure 1) for the two cases. It was observed that, at Y=3.2cm, due to the sudden expansion, the flow separated from the VF wall.

Therefore, high velocity gradient only concentrated at the core region of the jet. The velocity gradient profile was very similar for the two cases at this location. However, at $Y=4.0$ cm, the velocity gradient was much higher in (+)FVF due to the restriction of the channel. It indicated higher friction loss at this location in (+)FVF. Therefore, we propose that the final effect of FVFs on flow resistance is a combined effect of the reduced **mixing-related** minor loss and the increased friction-related major loss. Since both mechanisms are sensitive to the FVFs gap, it suggests an important role of the FVFs gap in determining the flow resistance. The hypothesis can also explain the experimental results in Agarwal et al.². For small FVFs gaps, the friction loss increased significantly and the increase may be more than the decrease in the mixing loss, thus, the flow resistance increased². For large FVFs gaps, the increase in the friction loss was small. At the same time, the FVFs had little effect on reducing the jet deflection and flow mixing. Therefore, the effect of FVFs on the flow resistance was very weak. For intermediate FVFs gaps, the increase in the friction loss and the decrease in the mixing loss were very close. Therefore, the flow resistance was observed to almost remain constant.

Flow resistance is related to laryngeal pressure drop. It is useful to examine the effect of the FVFs on laryngeal pressure distribution. Figure 7 shows the time-averaged pressure distribution along the flow direction at the centerline of the larynx for the two cases. It was observed that, with the addition of the FVFs, the pressure near the glottal exit decreased, and it stayed at a lower value through the ventricle and FVFs region. The pressure started to recover only after the FVFs. However, in (–)FVF, the pressure started to recover immediately after the glottis exit. The delayed recovery of the pressure in (+)FVF was because that the FVFs restricted the flow channel and therefore increased the flow velocity. As the Bernoulli effect is the dominant mechanism of the flow behavior in this region, the increased flow velocity decreased the pressure. Figure 7 also shows the time-averaged streamwise velocity distribution along the centerline of the larynx for the two cases. It confirmed that the flow velocity near the glottis exit was higher in (+)FVF. The slightly decreased pressure inside the glottis in (+)FVF may imply a decreased driving force of the VF motion. In fact, it was found that the maximum glottal opening decreased 3.29% with the addition of the FVFs, as shown in Table 2.

IV. Conclusion

A three-dimensional computational laryngeal model was used to study the effect of FVFs on laryngeal flow resistance. The results were discussed in a comparative manner with the previous 2D model⁴. On one hand, the results demonstrated the similar mechanism as observed in the 2D model that the presence of the FVFs suppressed the deflection of the glottal jet and in doing so, reduced the **mixing-related** minor loss in the supraglottal region. On the other hand, the 3D flow was more stable and straighter, so the effect of FVFs on suppressing the jet deflection in the 3D model was not as prominent as in the 2D model. Furthermore, the presence of the FVFs also increased the friction-related major loss due to the increased velocity gradient in the restricted flow channel. Therefore it was hypothesized that the final effect of the FVFs on the flow resistance is the combined effect of the reduced **mixing-related** minor loss and increased friction-related major loss. Since both effects are sensitive to the FVFs gap, this hypothesis suggests an important role of the FVFs gap in

determining flow resistance. It remains to be examined with systematic and quantitative studies in the future. The addition of FVFs was also found to decrease the pressure near the glottis exit and FVFs region.

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Figure 1.

(a) The computational domain and laryngeal model; (b) The mid-coronal plane of the laryngeal model; The dash lines and dots indicate the location of the measurements in Figure 6 and Figure 7; (c) The superior view of the geometrical model; (d) Dimensions of the FVF; (e) The three-layer inner structure of the TVF. Color online.

Figure 2.
Time history of the glottal volume flow rate of the two cases.

Figure 3. Frequency spectrum of the glottal flow rate over the sustained vibration stage of the two cases.

Figure 4. Contour of the velocity magnitude at the center plane of the larynx at phases 1, 3, 5, 7, 11 and 13. The phases are indicated in a typical glottal flow waveform. Color online.

Figure 5. Contour of the turbulent kinetic energy at the mid-coronal plane at the same phases as in Figure 4 for the two cases. Color online.

Figure 6.

The streamwise velocity of the time-averaged flow across the channel at two locations at $Y=3.2$ cm (near glottis exit) and $Y=4.0$ cm (near minimum FVFs gap) for the two cases. Color online.

Figure 7.
The time averaged pressure and velocity distribution along the flow direction at the centerline of the larynx for the two cases

Table 1

Material properties of the three layers of the VF. ρ is the tissue density, ν_p is the in-plane transversal Poisson's ratio, ν_{pz} is the longitudinal Poisson's ratio, E_p is the transversal Young's Modulus, E_{pz} is the longitudinal Young's Modulus, G_{pz} is the longitudinal shear modulus, and η is the damping coefficient.

	ρ	ν_p	ν_{pz}	E_p	E_{pz}	G_{pz}	η
	g/cm^3			kPa	kPa	kPa	$Poise$
Cover	1.043	0.9	0.0	2.041	20000	10	2
Ligament	1.043	0.9	0.0	3.306	33000	40	3.5
Body	1.043	0.9	0.0	3.99	40000	20	4

Table 2

Key flow quantities and their differences computed for the two cases and the differences of the corresponding quantities in the 2D model. F_0 is the fundamental frequency, Q_{peak} is the peak flow rate, $\bar{Q}$ is the time-mean flow rate, $\Delta P / \bar{Q}$ is the flow resistance, SNR is the signal-to-noise ratio, G_{peak} is the maximum glottal opening.

	F_0 (HZ)	Q_{peak} (ml/s)	$\bar{Q}$ (ml/s)	$\Delta P / \bar{Q}$ (Pa·s/ml)	SNR	G_{peak} (cm)
(-)FVF	237	291	191	5.24	19.1	0.0972
(+)FVF	243	299	197	5.08	35.6	0.0940
$\frac{(+)\text{FVF} - (-)\text{FVF}}{(-)\text{FVF}}$	+2.53%	2.75%	3.14%	-3.05%	+86.4%	-3.29%
2D Model	+2.2%	+47.4%	+21%	-17%	NA	