The Effect of False Vocal Folds on Laryngeal Flow Resistance in a Tubular Three-dimensional Computational Laryngeal Model

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Summary: Objective. The current study used a three-dimensional (3D) computational laryngeal model to investigate the effect of false vocal folds (FVFs) on laryngeal flow resistance.

Method. A 3D, 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 FVFs and the other without the FVFs, 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 (2D) computational model. On the 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 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 FVFs.

Key Words: False vocal folds–Computational laryngeal model–Laryngeal flow resistance–Glottal jet deflection–Three-dimensional glottal flow dynamics.

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 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 approaches.[1–4] 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 (FSI) computational laryngeal model.[4] 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 resistance.[5] 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 with the 2D flow.[6] 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 results[4] 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 an effect on the flow resistance. It needs to be pointed out that the distance between the TVFs and FVFs was around 0.6 cm in both models. The potential reason for this discrepancy could be the three-dimensionality of the flow, flow pulsatility or FSI, 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 ($Z_L = \Delta P_L / \bar{Q}$), the flow resistance for many cases in Alipour

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et al[3] was still increased with the presence of the FVFs (refer to table 2 and figures 3 and 5 of their paper). Therefore, the experimental observations in Alipour et al[3] also contradicted the observations in 2D computational models.

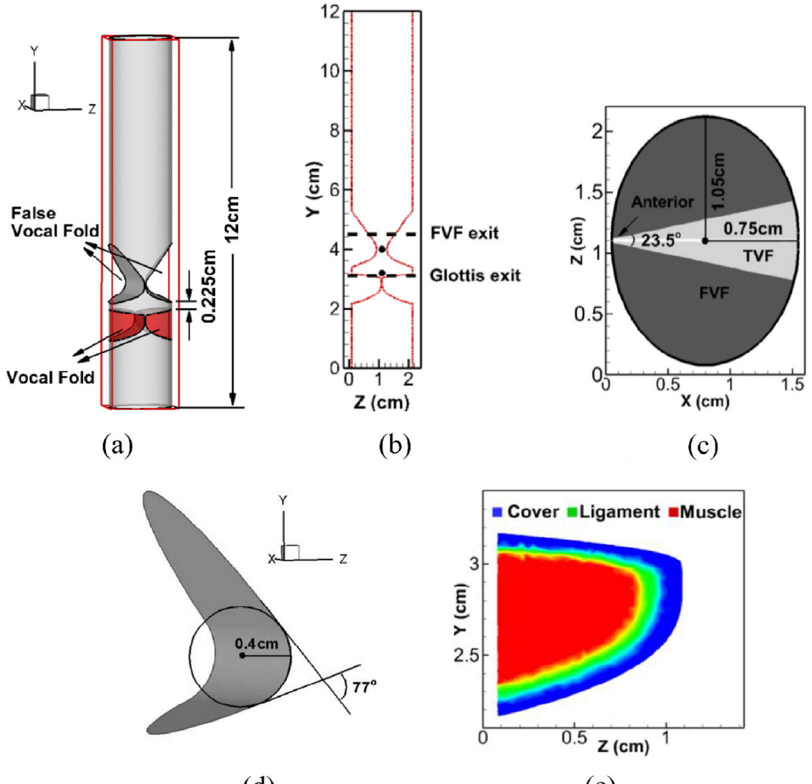
The discrepancies between the experimental studies and 2D computational studies resulted to the use of a 3D FSI computational laryngeal model to examine the effect of FVFs on laryngeal flow resistance in the current study. A 3D, tubular-shaped laryngeal model that has been designed with a high level of realism with respect to the human laryngeal anatomy was utilized. As an extension of the previous 2D computational model,[4] the results were extensively compared to the 2D model.

SIMULATION SETUP

The numerical algorithm and simulation setup have been reported in detail in our previous studies.[7,8] 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 used an explicitly coupled immersed-boundary-finite-element method-based FSI solver to model human phonation.[8] 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 1A. 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 1.6 cm × 12 cm × 2.2 cm rectangular computational domain. The cross-section profile of the TVF was generated based on a high-resolution laryngeal computed tomography scan of a normal male subject.[7] A 0.01-cm 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* measurements[9] (Figure 1D). Figure 1C is the superior view of the geometrical model of the vocal tract. The pair of FVFs was placed into the elliptical vocal tract with an anterior angle of 23.5°, which was measured from the laryngealscopic superior-to-inferior view of the larynx of a human subject.[8] 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 $y = 3.39$ cm 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.



(a) (b) (c)

(d) (e)

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 Figures 6 and 7; **C.** The superior view of the geometrical model; **D.** Dimensions of the FVF; **E.** The three-layer inner structure of the TVF. (For interpretation of the references to color in this figure, the reader is referred to the Web version of this article.)