Characterizing Liquid Redistribution in a Biphasic Vibrating Vocal Fold Using Finite Element Analysis

Anton A. Kvit, Erin E. Devine, Jack J. Jiang, Andrew C. Vamos, and Chao Tao, *Madison, Wisconsin*

Summary: Objectives. Vocal fold tissue is biphasic and consists of a solid extracellular matrix skeleton swelled with interstitial fluid. Interactions between the liquid and solid impact the material properties and stress response of the tissue. The objective of this study was to model the movement of liquid during vocal fold vibration and to estimate the volume of liquid accumulation and stress experienced by the tissue near the anterior-posterior midline, where benign lesions are observed to form.

Methods. A three-dimensional biphasic finite element model of a single vocal fold was built to solve for the liquid velocity, pore pressure, and von Mises stress during and just after vibration using the commercial finite element software COMSOL Multiphysics (Version 4.3a, 2013, Structural Mechanics and Subsurface Flow Modules). Vibration was induced by applying direct load pressures to the subglottal and intraglottal surfaces. Pressure ranges, frequency, and material parameters were chosen based on those reported in the literature. Postprocessing included liquid velocity, pore pressure, and von Mises stress calculations as well as the frequency-stress and amplitude-stress relationships.

Results. Resulting time-averaged velocity vectors during vibration indicated liquid movement toward the midline of the fold, as well as upward movement in the inferior-superior direction. Pore pressure and von Misses stresses were higher in this region just after vibration. A linear relationship was found between the amplitude and pore pressure, whereas a nonlinear relationship was found between the frequency and pore pressure.

Conclusions. Although this study had certain computational simplifications, it is the first biphasic finite element model to use a realistic geometry and demonstrate the ability to characterize liquid movement due to vibration. Results indicate that there is a significant amount of liquid that accumulates at the midline; however, the role of this accumulation still requires investigation. Further investigation of these mechanical factors may lend insight into the mechanism of benign lesion formation.

Key Words: Finite element analysis–Vocal folds–Biomechanics–Benign lesions.

INTRODUCTION

Like many tissues in the body, the vocal folds are most accurately described as a biphasic material—consisting of both porous solid skeleton composed of proteins, carbohydrates, lipids, collagen, and elastin fibers and liquid filling the space between the solid components.[1] The porosity of the solid skeleton is measured as the void space, which in a biphasic tissue is filled with interstitial fluid. The interaction between the solid matrix and the interstitial fluid is thought to affect pressure, stress, and strain distribution within the vocal fold during vibration.[2] Other important sources of mechanical stress include deformation of the vocal fold tissue due to the air pressure driving phonation and the contact stress that occurs during vocal fold collision. The combination of these mechanical forces is believed to stress the extracellular matrix and affect the pressures experienced by capillaries, potentially leading to fiber damage and capillary rupture,[3] resulting in tissue remodeling and subsequent formation of benign lesions.[4] Although the interplay between all of the listed forces has yet to be thoroughly explored, currently particularly lacking in literature is an investigation of the mechanical effects of liquid redistribution, which only recently has been explored using computer modeling.[2,5,6]

Mathematical modeling has been used extensively to obtain information on vocal fold oscillation, collision, and biphasic behavior[7–9] because it allows researchers to easily vary parameters and take measurements that are difficult to attain experimentally. The earliest models consisted of discrete mechanical oscillators connected by springs and dampers, which allowed easy numerical solutions, but did not reflect the true geometry or composition of the vocal folds.[10] Continuous models that viewed vocal fold tissue as viscoelastic solids, in which the tissue is modeled as having both viscous and elastic properties, were developed, producing more elaborate vibration patterns and allowing the addition of the mucosal wave.[11] Alternatively, Tsai et al[12] produced a model which viewed the fold entirely as a liquid. The limitation of all the listed models is that they assume the vocal fold material to be a single phase, as opposed to a biphasic entity. To study the interaction between the solid and liquid phases, models using a finite difference method were developed that viewed the vocal fold as a transversely isotropic poroviscoelastic solid, where the tissue is modeled as a porous material with both viscous and elastic properties, with liquid-filled pores.[2,5,6] This showed liquid accumulation in the anterior-posterior midline of the vocal fold in amounts proportional to frequency and amplitude

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From the Division of Otolaryngology – Head and Neck Surgery, Department of Surgery, University of Wisconsin-Madison School of Medicine and Public Health, Madison, Wisconsin.

Address correspondence and reprint requests to Jack J. Jiang, Division of Otolaryngology – Head and Neck Surgery, Department of Surgery, University of Wisconsin-Madison School of Medicine and Public Health, 1300 University Avenue, 5745 Medical Sciences Center, Madison, WI 53706. E-mail: jjjiang@wisc.edu

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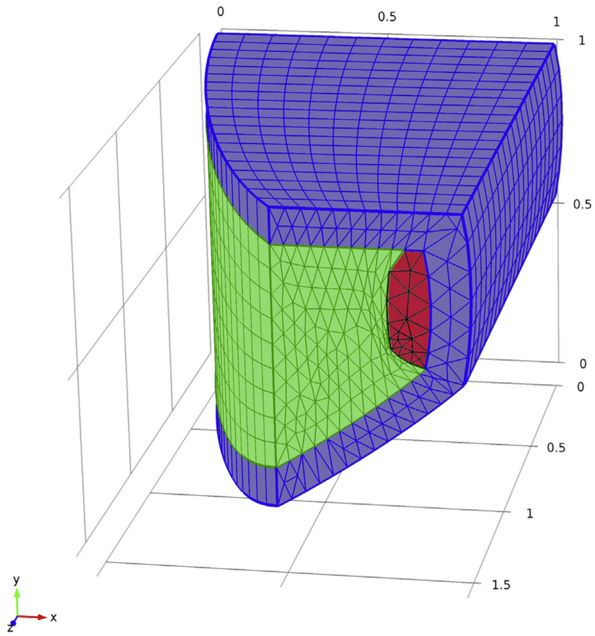

FIGURE 1. The finite element model including the cover, ligament, and body layers with a curved lateral surface.

of oscillation.[5] This midline accumulation, and subsequent rise in liquid pressure, has been previously observed in a mechanical model of a vibrating capillary[3] and is thought to contribute to benign vocal fold injury development.[3,13]

Although biphasic vocal fold tissue has been modeled as a bunch of one-dimensional fibers[5] and a three-dimensional rectangular parallelepiped,[2] no such models reflect the geometry and layering of the fold, the pressures experienced by the subglottal vocal fold surface, or the elliptical vibration trajectory, which vocal folds have been shown to follow.[7,14] Finite element analysis (FEA) allows the inclusion of complex three-dimensional geometries and boundaries in mathematical models and has been previously used for vocal fold representation, allowing for complex numerical solutions,[7–9] always assuming, however, single-phase material properties. Thus, no model so far has included both realistic geometry and tissue porosity, which is necessary to accurately investigate the mechanical effects of vocal fold vibration and liquid redistribution on pathology development.

The proposed model was built and analyzed using a commercial FEA software, which combines these two parameters, and therefore more accurately represents the true nature of the vocal folds. In this representation, interporous liquid redistribution is affected not only by varying intraglottal pressure, amplitude, and frequency, as in previous models[2,6] but also by varying pressure applied on the subglottal surface, elliptical oscillation trajectory, complex vocal fold geometry, and its various layers. This analysis permits modeling deformation of the elastic solid skeleton as the result of liquid accumulation and spatiotemporal distribution of solid and liquid pressures. Furthermore, it provides a basis on which more complex analyses including quantifying stress damage and pathology modeling could be implemented.

METHODS

Model development

Geometry. A three-dimensional finite element model was defined in Cartesian coordinates, where the *x*-dimension was the lateral direction from glottal midline, the *y*-dimension defined the vocal fold in the inferior-superior direction, and the *z*-dimension defined the anterior-posterior distance. The length (*z*-direction) of the vocal fold was 1.6 cm, whereas the thickness (*y*-direction) was 0.45 cm, similar to previous finite element models[7,9] and within the human physiological range measured by Hirano and Sato.[15] The depth (*x*-direction) of the vocal fold decreased from 1.0 cm on the posterior side to 0.5 cm on the anterior side, following the formula: $\text{Depth}(z) = 1.0 - 0.5(z/\text{Length})^2$, devised by Titze and Talkin.[16] The geometry can be seen in Figure 1.

Material properties. The solid skeleton was modeled as three poroelastic transversely isotropic layers, defined as having both porous and linearly elastic material properties,[15] representing the body, ligament, and cover of the vocal fold. Linear elasticity was an approximation because the strains observed in the model fall within the linear region.[17] Transverse Young's modulus of the body was 4 kPa, and transverse Young's moduli of the cover and ligament were 2 kPa, according to physiological measurements[18] and previous models.[8,9] The longitudinal shear moduli for the cover, ligament, and body were 10, 40, and 12 kPa, respectively, based on previous measurements and models.[9,19,20] The longitudinal Young's moduli for the cover, ligament, and body were set to be 40, 30, and 25 kPa based on previous physiological measurements and models.[9,17,21] The interporous liquid was defined as an incompressible viscous fluid estimated to have properties of water. The vocal fold was modeled as nearly incompressible; an assumption that has been used in numerous previous models[22–24]; and therefore, the Poisson's ratio was set to be 0.499 for all layers in all directions.

Darcy's law governs the movement of fluid through a porous medium. Darcy flux is defined as follows:

$$\boldsymbol{u}_f = -\frac{\kappa}{\mu}\nabla p$$

where $\boldsymbol{u}_f$ (meters per second) is the flux across a porous surface, κ (square meters) is the permeability of the porous medium, μ (Pa*s) is the fluid's dynamic viscosity, and p (Pa) is the fluid's pressure (COMSOL Multiphysics Version 4.3a, 2013, COMSOL Inc. Burlington, MA).

Porosity, a measure of void space in a material, was set to 0.8 based on dehydration studies of the vocal folds.[25] It has been shown that the vast majority of fibers in vocal fold tissue are aligned longitudinally,[26] and the permeability of liquid parallel to the fibers is significantly larger than in the transverse directions.[6] Therefore, the longitudinal permeability in the model was set to be 6e-13 m^2, and 2e-13 m^2 in all other directions, which is within the range measured in a bovine vocal fold scaffold.[27] Table 1 summarizes material parameters.

Boundary conditions. The vocal fold was constrained on the lateral, posterior, and anterior faces, whereas the superior