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Characterizing liquid redistribution in a biphasic vibrating vocal fold using finite element analysis

Anton A. Kvit, Erin E. Devine, MS, Andrew C. Vámos, Chao Tao, PhD, and Jack J. Jiang, MD, PhD

University of Wisconsin-Madison School of Medicine and Public Health, Department of Surgery, Division of Otolaryngology – Head and Neck Surgery, Madison, WI, 53792

Abstract

OBJECTIVE—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 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. Post-processing 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 towards the midline of the fold, as upwards movement in the inferior-superior direction. Pore pressure and von Mises stresses were higher in this region just following vibration. A linear relationship was found between the amplitude and pore pressure, while a nonlinear relationship was found between the frequency and pore pressure.

CONCLUSIONS—While this study had certain computational simplifications, it is the first biphasic finite element model to employ a realistic geometry and demonstrated the ability to characterize liquid movement due to vibration. Results indicate that there is a significant amount

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Corresponding author: Jack Jiang, M.D., Ph.D., 1300 University Avenue, 5745 Medical Sciences Center, Madison, WI 53706, Telephone: 608-265-9854, Fax: 608-265-2139, jjjiang@wisc.edu.

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Conflict of interest statement

None of the authors have conflicts of interest to report.

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.

Keywords

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 a 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 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). While 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) since 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, 2006 (12) produced a model which viewed the fold entirely as a liquid. The limitation of all of the listed models is that they assume the vocal fold material to be a single phase, as opposed to a biphasic, entity. In order 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) and showed liquid accumulation in the anterior-posterior midline of the fold in amounts proportional to frequency and amplitude 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).

While biphasic vocal fold tissue has been modeled as a bunch of 1-D fibers (5) and a 3-D rectangular parallelepiped (2), no such models reflect the geometry and layering of the fold, the pressures experienced by the subglottal 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 3-D 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, built and analyzed using a commercial FEA software 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, as well as 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 3-D finite element model was defined in Cartesian coordinates, where the x-dimension was the lateral direction from glottal midline, the y-dimension defined the 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, while 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, 1993 (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 $Depth(z) = 1.0 - 0.5(z/Length)^2$, devised by Titze and Talkin, 1979 (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 since 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) as well as previous models (8, 9). The longitudinal shear moduli for the cover, ligament and body were 10kPa, 40kPa, 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 40kPa, 30kPa, and 25kPa, 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 utilized 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:

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

where $\mathbf{u}_f$ (m/s) is the flux across a porous surface, κ (m²) 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).

Porosity, a measure of void space in a material, was set to 0.8 based on dehydration studies of the vocal folds (Hanson, 2010). It has been shown that the vast majority of fibers in vocal fold tissue are aligned longitudinally (25), 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 m², and 2e–13 m² in all other directions, which is within the range measured in a bovine vocal fold scaffold (26). Table 1 summarizes material parameters.

Boundary Conditions—The vocal fold was constrained on the lateral, posterior and anterior faces, while the superior surface had a free boundary condition, and all the surfaces were set to be impermeable to liquid. Tao et al., 2006 (9) identified three factors contributing to vocal fold closure in his self-oscillating model, which include the elastic recoil of the deformed tissue, decreased intraglottal pressure due to airstream velocity, and a decrease in subglottal pressure. While the current model does not utilize self-oscillation to drive and sustain phonation, the same three factors were implemented to induce vocal fold closure. Similar to a model by Gunter, 2003 (27), boundary loads in the form of distributed pressure were applied on the subglottal and intraglottal surfaces of the fold during opening. The elastic force of the vocal fold, along with negative pressure along the intraglottal surface (9, 28) caused vocal fold closure. Subglottal and intraglottal surface pressures were applied directly to the subglottal and intraglottal surfaces and varied sinusoidally with time and had a phase difference of 90 degrees in order to create a physiologically viable elliptical vibration pattern. A minimum subglottal pressure of 0.4 kPa was maintained throughout vibration. Tao et al., 2010 (5) saw higher liquid accumulation at higher vibration amplitudes, however did not translate these findings into stresses on the solid matrix. In order to observe the effect of vibration amplitude on liquid accumulation and subsequent stresses, the fold was vibrated at approximately 0.6 mm, 0.8 mm, 1.0 mm, 1.2 mm, and 1.4 mm amplitudes. The intraglottal and subglottal pressure ranges utilized in order to achieve each amplitude, described in Table 2, are close to physiologically measured values (29, 30) and are comparable in magnitude to pressures utilized in previous models to achieve similar oscillation (2, 5, 8). Oscillation frequency was kept constant at 200 Hz for each amplitude. This frequency was chosen since it is within the normal vocal range for adult women (31), a

population in which vocal nodules are more likely to occur (32). Furthermore, accumulation has been demonstrated to be greater at higher frequencies in a model utilizing one-dimensional biphasic fibers (5). In order to determine the effect of frequency on liquid redistribution in three dimensions and quantify the resulting structural stresses, the fold, held at a constant 0.8mm vibration amplitude, was oscillated at 50 Hz, 100 Hz, 200 Hz, 400 Hz, and 1000 Hz frequencies.

The vocal fold in all instances oscillated for 0.2 seconds, allowing liquid to accumulate at the midline to the point where the midline was fully saturated. This time was selected from pilot runs of the model. A step function was applied at 0.2 seconds, rapidly decreasing both the intraglottal and subglottal pressures at the same rate, and oscillation stopped entirely at 0.204 seconds, returning the tissue to its original position. The fold was then allowed to rest for 0.1 more seconds to allow the accumulated liquid to flow back to the anterior-posterior periphery, in order to reveal the extent of bulging and pressure caused by liquid accumulation alone, without the effects of vocal fold displacement. A single fold was modeled, and contact pressure was ignored as in previous biphasic models (2, 5, 6).

The model was built, converged, and the solution was obtained and analyzed using COMSOL Multiphysics (Version 4.3a, 2013, Structural Mechanics and Subsurface Flow Modules) commercial finite element software. The 3D fold was meshed using 2640 prism-shaped elements, and was solved for 39,153 degrees of freedom. In order to properly capture the details of vibration, time steps with a length of 0.2 ms were used.

Post Processing

Trajectories—Figure 2 features a transverse view of the 1.4 mm vibration amplitude vocal fold model at the anterior-posterior midline, along with trajectory traces of a single complete oscillation at seven nodes. The trajectories took varying elliptical shapes, with highest total displacement near the superior medial surface, where the ellipsis is slanted in the x-direction, and lower total displacement near the inferior surface, where the y-direction is more prevalent. This trajectory pattern mimics previous modeling and excised results (7, 14, 33) in models with similar vibration amplitudes, and displacement magnitudes are comparable to ones observed physiologically (34). Additionally the movement of the vocal folds and mucosal wave can be seen in Figure 3.

A number of previous vibrating vocal fold models have included self oscillation, which results from coupling of airflow to vocal fold movement (7–9, 35), and provides a physiologically relevant way to set the folds in motion. Utilizing this method allows researchers to explore the effects of airflow aerodynamics, as well as vary boundary loads not only with time, but also across the surface, allowing visualization of the mucosal wave (9). Self-oscillation can be beneficial, however it is computationally costly, and liquid accumulation in the anterior-posterior midline has been observed in a far more robust mechanical model (3), which ignored not only the mucosal wave, but also incorporated vibration in only a single direction. Thus self-oscillation was not necessary for the current model, which aimed to qualitatively access the effect of vocal fold geometry and varying subglottal pressure on liquid redistribution and provide preliminary estimates of the resulting pore pressures and stresses. Due to the high computational demands of modeling interaction

of the airflow, the poroelastic solid skeleton and the interporous liquid, boundary loads in the current model have been applied directly, reducing convergence speed without sacrificing the qualitative outcome.

Interporous liquid movement—Liquid redistribution in the vocal fold was evaluated in three dimensions. Interporous liquid velocity was time-averaged to evaluate the net movement of interporous liquid over the course of vibration. Interporous liquid movement is represented as scaled vectors distributed throughout the geometry. Interporous liquid velocity is the velocity of the fluid occupying the pores of the solid skeleton of the vocal fold tissue, derived by dividing the Darcy flux by porosity.

Structural stress and pore pressure—The pressure exerted by the interporous liquid, referred to as pore pressure, was mapped in three dimensions 0.01 seconds after vibration stopped. Furthermore, the maximum pore pressure ($P_{\max}$) and maximum von Mises stress ($\sigma_{\max}$) during vibration as well as the change in pore pressure (P) and von Mises stress (σ) due only to liquid accumulation, were recorded at a point in the anterior-posterior midline for each vibration amplitude. The coordinates (0.95cm, 0.95cm, 0.8cm) were chosen for the point since vocal nodules are observed to form in the superior aspect of the medial fold surface, and liquid accumulation was shown to occur there (2, 6), possibly contributing to vocal pathology development (3). Von Mises stress is a scalar quantity used to represent composite stress, and was recorded since it can help determine the damage-inducing threshold of forces experienced by the vocal fold due to both liquid accumulation and vibration. It has been measured in a previous model (27) that also sought to quantify stress experienced by the vocal folds. In order to separate the stresses and pressures that resulted from liquid accumulation, the peak stresses and pressures at the start of vibration (where no liquid accumulation has yet occurred) were subtracted from the maximum observed structural stresses and pressures after liquid redistribution stabilized.

Results

Representation of interporous liquid velocity vectors from the 0.8 mm vibration amplitude model averaged over time during oscillation (Figure 4) depicts liquid redistribution in the superior-inferior direction (x-y plane). Figure 5 represents the fluid redistribution in the anterior-posterior direction (x-z plane). Red arrows represent velocity vectors and direction of interporous liquid movement at 343 different locations throughout the fold. Averaged between 0 and 0.2 seconds, these vectors show net midline accumulation near the superior and intraglottal surfaces, however this trend is somewhat obscured by the robust movement from the inferior to the superior surface. Nevertheless, this upwards trend does not affect the relative longitudinal liquid accumulation, which is apparent at the medial surface when a plane of velocity vectors 1mm below the superior surface of the vocal fold is viewed from above (9).

Figure 6 depicts the bulging of the vocal fold midline at 0.204 seconds, the instant the oscillation stops completely in the 0.8mm vibration amplitude model. The bulging is scaled 100 times the original size for ease of visualization. Pore pressure distribution, depicted as well using a color gradient, is indicative of interporous liquid accumulation within the vocal

fold which peaks at the anterior-posterior midline, since liquid within a vibrating vessel moves towards the point of highest vibratory amplitude (3). Von Mises stress distribution follows a similar trend as the pore-pressure, increasing at the anterior-posterior midline. The pattern is somewhat obscured, however, by bands of high pressure lying transverse to the vocal fold at the anterior-posterior peripheries. These bands are likely caused by edge effects within the finite element software, and are not expected to be observed physiologically (Figure 6).

$\sigma_{\max}$, $P_{\max}$, P , and σ , were recorded at the anterior-posterior midline one mm under the surface of the vocal fold for each of the five vibration amplitudes (Table 3). Additionally, $P_{\max}$ at each vibration amplitude (Figure 8a) as well as $\sigma_{\max}$ at each vibration amplitude (Figure 8b) were plotted, revealing that in this model, pore pressure and von Mises stress both increase linearly with amplitude. Plotting P against varying oscillation frequencies revealed a positive non-linear relationship between the two parameters (Figure 9).

Discussion and Conclusions

The goal of the present model was to simulate interporous liquid redistribution during vocal fold vibration and characterize the resulting interporous pressure and solid stresses. This was accomplished by obtaining time averaged liquid velocity vectors, pore pressure and stress distributions, as well as deformation of vocal fold tissue from the analysis of our results.

Time averaged velocity vectors (Figure 4) depict net liquid movement towards the superior aspect of the anterior-posterior midline of the vocal fold, as was expected. However, the vectors also show net movement of interporous liquid from the inferior to the superior aspect of the vocal fold. This displacement is likely caused by the varying subglottal pressure applied to the inferior surface of the vocal folds, while the supraglottal pressure remains atmospheric (9), and therefore receives no external boundary loads in the model. A similar pattern of liquid displacement was noted by Tao et al., 2013 (6), who hypothesized that upwards liquid movement was caused by mucosal wave propagation. That model, however, as well as all previous biphasic models, ignored the effects of subglottal pressure on the inferior surface, since the vocal folds were assumed to have parallelepiped geometry (5, 6). It is therefore possible that liquid movement in the inferior-superior direction is due to a combination of the mucosal wave and subglottal pressure. Inertial effects, similar to the ones responsible for anterior-posterior accumulation during vibration (3), are likely also responsible for the observed lateral-medial liquid displacement (Figure 5). Finally, it is possible that the boundary conditions exaggerated the inferior-superior liquid movement somewhat due to the differences between subglottal pressure and atmospheric pressure and the lack of restriction of liquid movement in the body of the vocal folds. Due to the difficulty of measuring physiological liquid redistribution in a physical vocal fold model, these findings cannot currently be validated. However, based on our modeling results, it is possible that in addition to anterior-posterior liquid accumulation, accumulation in the inferior-superior and lateral-medial directions contributes to increased pore pressure and structural stress in the superior-medial region of the vocal fold. These combined pressures could subsequently contribute to vocal nodule or other pathology development (3, 13).

The interporous liquid accumulation that is evident from velocity vector analysis is also reflected in the pore pressure distribution of the vocal fold immediately following vibration. Visualizing pore pressure distribution at the end of vibration (Figure 6) revealed areas of high pressure, as well as vocal fold bulging, in the anterior-posterior midline and along the lateral subglottal surface. The pattern of von Mises stress distribution was very similar to the observed pore pressure distribution (Figure 7), suggesting that areas of the solid matrix that undergo liquid accumulation experience higher stress, that, in combination with vibratory and collision stresses, could lead to damage and subsequent pathology formation.

The observed ($\sigma_{\max}$) during vocal fold vibration for the maximum amplitude of 1.4 mm was 2707.4 Pa. The resulting (σ) was 148.4 Pa. While this stress is small in relation to ($\sigma_{\max}$), it may still play a significant role in the development of vocal fold lesions when combined with the stresses of tissue deformation and contact pressure. On the other hand, the fluid accumulation may actually act to cushion the solid matrix during vocal fold collision by absorbing some of the impact stress (6). However, the effect of pore pressure changes on the vasculature have not yet been extensively studied, and may lend insight into capillary damage and edema during vocal fold vibration that may contribute to the remodeling of the vocal folds, leading to vocal fold nodules and other benign lesions (4). This preliminary study into the fluid dynamics during vocal fold vibration of healthy tissue allows us to begin to make hypotheses, however, further study into the fluid dynamics in pathological vocal fold vibration are necessary before drawing conclusions as to the role that fluid accumulation plays. Kuebler, 2002 (36) found that pressures as low as 500 Pa can trigger inflammatory pathways in the vasculature through Ca^{2+} dependent signaling, and $P_{\max}$ observed in the model surpasses that value even at lower vibration amplitudes at 200 Hz frequency (Table 3). Future work might focus on the effects that the mechanical environment created by this model would have on both capillaries and connective fibers like collagen and elastin in the vocal folds at the superior midline. This would lend insight into whether the mechanism of lesion formation is due to inflation caused by mechanical damage of connective tissues, or rather, inflammation from capillary swelling and damage.

The model presented in this study satisfactorily accomplishes the outlined qualitative goals; however it has several limitations that could be addressed in the future in order to more accurately estimate the stresses seen in the system. The limitations are introduced by the incomplete parameterization of vocal fold tissues (37). Thus, many assumptions are made in order to simplify computation or fill gaps in knowledge that currently limit the clinical applicability of the model. As imaging technologies and rheological techniques improve, more accurate geometries and parameterization can be implemented. The encouraging results of our model suggest that pursuing these lines of research would greatly benefit the investigation of vocal fold effects on pathology formation, decreasing the necessity of animal experimentation.

The approximation of our solid matrix as linearly elastic was used to reduce the computational costs of the model. Tao, 2010 (5) observed that pore pressure, and subsequently structural stress, was quadratically dependent on amplitude and frequency; however, that model incorporated self-oscillation, which makes analyzing frequency separate from amplitude impossible. Since we applied varying boundary loads, and therefore

frequency, directly, we were able to analyze the two parameters independently. We found a linear dependence between vibration amplitude and pore pressure, which is likely due to the fact that the presented model is perfectly elastic, while the Tao, 2010 model was viscoelastic. The relationship between frequency and pore pressure, on the other hand, was non-linear and less dramatic (Figure 9), suggesting that vocal fold vibration frequency has a less significant effect on liquid redistribution than vibration amplitude, and is independent of material viscoelasticity. It is known that vocal folds have are not perfectly elastic, but rather viscoelastic with a linearly elastic region (38), and therefore can be permanently deformed at high enough stresses (17). A linear elastic approximation is valid for the observed deformation, and has worked well for previous models (8, 27). However, viscoelastic material parameters might be used in future models to better approximate the tissue response to stress. Contact pressure could be added to the model to study the stresses and strains as well as liquid redistribution resulting from the presence of another fold and to determine the role of interporous liquid in stress absorption during collision.

The presented model is the first finite element vocal fold representation that incorporates complex vocal fold geometry and varying subglottal pressure along with a poroelastic material, showing how the interaction of these parameters affects liquid redistribution. The model allows the visualization of interporous liquid distribution in three dimensions, as well as vocal fold bulging that occurs due to liquid accumulation and pore pressure increase, and might contribute to nodule or edema formation. Pore pressure and von Mises stress were shown to increase with vibration amplitude as well as frequency, suggesting that more damage occurs with yelling, high pitch vocalizing, or other strenuous vocal exercise. More complex models could be built on the basis of the presented one that could help quantify the forces responsible for pathology development.

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Figure 1.
The utilized finite element model including the cover, ligament, and body layers as well as a curved lateral surface.

Figure 2.

Trajectories of a single vocal fold oscillation at seven varying nodes of the 1.4mm vibration amplitude model at the maximum amplitude transverse plane (0.80mm). The pattern is similar to the one observed in a previous models with a similar vibration amplitudes (Alipour et al., 2000; Berry et al., 2001).

Figure 3.
Depiction of the frames of vocal fold movement over time during an oscillation cycle.

Figure 4.

Interporous liquid time averaged velocity vectors during oscillation (0 – 0.2 s) depict anterior-posterior midline accumulation near the superior surface, along with inferior-superior liquid redistribution throughout the entire vocal fold.

Figure 5.

Time averaged plane of velocity vectors 1mm below the superior surface of the vocal fold during oscillation (0–0.2s) reveals anterior-posterior midline accumulation at the medial surface, as well as lateral-medial liquid redistribution.

Figure 6.
Bulging and high pore pressure in the superior-medial aspect of the vocal fold due to accumulated liquid at 0.204 second instant, immediately after vibration stopped.

Figure 7.

Von Mises stress distribution felt by the solid matrix at the 0.204 second instant, immediately after vibration stopped. Medial-lateral high-stress bands at the periphery of the fold are likely due to edge effects, and are not expected to occur physiologically.

A)

B)

Figure 8.

A: Maximum structural stress ($\sigma_{\max}$) during vibration experienced at the anterior-posterior midline of the vocal fold at 0.6, 0.8, 1.0, 1.2, and 1.4 vibration amplitudes. B: Structural stress due only to liquid accumulation (σ) experienced at point B in the anterior-posterior midline of the vocal fold at 0.6, 0.8, 1.0, 1.2, and 1.4 vibration amplitudes.

Change in Pores Pressure v. Frequency

Figure 9.
 $P_{\max}$ during vibration experienced at the anterior-posterior midline of the vocal fold at 50, 100, 200, 400 and 1000 Hz vibration amplitudes.

Table 1

Input parameters used in the model

Porosity	0.8
Liquid Density	1000kg/m ³
Liquid Compressibility	0
Dynamic liquid viscosity	8.9×10^{-4} Pa*s
Longitudinal permeability	6e-13 m ²
Transverse permeability	2e-13 m ²

Table 2

Intraglottal and subglottal pressure ranges

Vibration Amplitude:	0.6mm	0.8mm	1.0mm	1.2mm	1.4mm
Subglottal Pressure Range (Pa)	0–950	0–1310	0–1500	0–1900	0–2160
Intraglottal Pressure Range (Pa)	–237.5–712.5	–327.5–982.5	–375–1125	–475–1425	–540–1620

Table 3

Von Mises stress and pore pressure at the maximum amplitude point

Vibration amplitude	$\sigma_{\max}$ Maximum von Mises Stress (Pa)	$P_{\max}$ Maximum Pore Pressure (Pa)	σ von Mises Stress Change (Pa)	P Pore Pressure Change (Pa)
0.6 mm	1192.5	477.5	66.5	173.3
0.8 mm	1641.0	658.5	91.5	238.9
1.0 mm	1881.7	754.2	106.4	274.1
1.2 mm	2382.5	957.3	132.1	348.7
1.4 mm	2707.4	1086.6	148.4	395.3