Simulation of vibration of the human vocal folds

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ABSTRACT: Three-dimensional (3D) finite element (FE) fully parametric model of the human larynx was developed and used for numerical simulation of stresses during vibrating vocal folds with collisions. The complex model consists of the vocal folds, arytenoids, thyroid and cricoid cartilages. The vocal fold tissue is modeled as a three layered transversal isotropic material. First, the basic frequency-modal properties of the model are presented for a given pre-stress of the vocal folds. Then the results of numerical simulation of the vocal folds oscillations excited by a prescribed intraglottal aerodynamic pressure are presented. The FE contact elements are used for modelling the vocal folds collisions and the stresses in the vocal fold tissue are computed in time domain. The results show significant dynamic stresses in all three directions (horizontal, vertical and anterior-posterior) and similar maxima of stresses were obtained for closed and open phase of the vocal folds motion. The values of computed stresses are comparable with the impact stress measured on the developed physical vocal folds model made of silicon rubber.

KEY WORDS: biomechanics of human voice, FEM, impact stress

1 INTRODUCTION

Voice problems are common, especially among professional voice users like teachers, actors and singers. The main reason can be fatigue due to the mechanical loading of the vocal fold tissue during voice production. For example, symptoms of vocal fatigue reflecting type of voice production and the effects of vocal loading were studied by Laukkonen et al. (2008) and consequences of the mucosal waves travelling on the vibrating vocal folds surface were discussed from the point of view of mechanical stress in phonation and the formation of vocal fold traumas by Sonninen and Laukkonen (2003).

Designing a model of the human vocal folds enabling to model some pathological situations and voice disorders is becoming an important part of voice research and such computer models are becoming also applicable in phonosurgery (Mittal et al. 2011, Chen T. et al. 2011). The mechanical loading is caused by a combination of the aerodynamic, inertial and impact forces during vocal fold self-oscillations with collisions. And, moreover, regarding the complicated three-dimensional (3D) structure and material properties of the living tissue, many unknown mechanisms for tissue damage remain to be investigated and explained. Mechanical stress is always encountered in phonation. This includes tensile stress, shear stress, impact stress during collision, inertial stress, and stress caused by aerodynamic pressure (Titze, 1994). Excessive collision and acceleration may be responsible for most tissue damage. Some simplified lumped mass dynamic models of phonation can be used for a rough estimate of the impact stress or acceleration level depending on various phonation parameters like e.g. prephonatory glottal gap, subglottal pressure and fundamental frequency (Horálek et al., 2005, 2007). Maximum impact stress about 4 kPa resulted from the aeroelastic model of vocal folds self-oscillations numerically simulated by Horálek et al. (2009).

However, more sophisticated models based on Finite Element (FE) modelling enable us to estimate all main normal and shear stresses in the different vocal fold tissue layers in all three directions, even if the computational demands on computers and computer time needed are much higher and still limited. The first FE models of the vocal fold vibration were developed from the basic laws of continuum mechanics to obtain the oscillatory characteristics of the vocal folds (see e.g. Alipour et al., 2003; Titze, 2006). A three-dimensional model was developed by Rosa et al. (2003) to simulate the larynx during vocalization. The FE method was used to calculate the airflow velocity and pressure along the larynx as well as tissue displacements. Following the hypothesis that vocal fold tissue lesions such as nodules and polyps are developed in response to mechanical stress occurring during vocal fold collision Gunter (2003, 2004) designed a 3D FE model of the vocal folds for predicting the entire stress tensor. The hypothesis was supported by predictions from the FE model that three components of normal stress and one component of shear stress are increased during collision in the typical locations of lesions. Given that stress estimation during phonation in the vocal folds is helpful in understanding vocal traumas, a self-oscillating FE model was developed by Tao et al. (2006) and Tao and Jiang (2007). They used the model for simulating vocal fold vibration during phonation. The spatial and temporal characteristics of mechanical stress in the vocal folds were predicted by this model. Spatially, the computed normal stress was significantly higher on the vocal fold surface than inside of the vocal folds. The peak-to-peak amplitude of the normal stress reaches its maximum value about 6 kPa at the midpoint of the medial surface for the subglottal pressure 2.5 kPa. In the inferior-superior direction
the maximum impact pressure was related to the narrowest glottis. Maximum impact stress 5.34 kPa was measured in the canine excised larynges by Verdolini et al. (1998, 1999) and 5 kPa in humans by Reed et al. (1992). Recently, three methods of the contact pressure measurements were tested on a silicone replica of human vocal folds by Chen and Mongeau (2011) during the self-oscillations. The maximum impact stress measured by a probe microphone was found 0.59 kPa for the subglottal pressure 2.84 kPa, the airflow rate 0.72 l/s and the vibration frequency 140 Hz.

With the intention to estimate the stresses in the vocal fold tissue during phonation, a 3D FE fully parametrical model of the larynx was developed. The model considers the phonation position of the vocal folds given by the possible independent motion of the cricoid, thyroid and arytenoid cartilages, i.e. enables uncomplicated variation of the geometrical configuration, the changes of the longitudinal tension (pre-stress) and abduction of the vocal folds as well as the shape and the nonlinear material properties of the individual vocal fold tissue layers. First, the frequency-modal properties of the model are presented. Then the results of the numerical simulation of the vocal folds oscillations excited by a prescribed periodic intraglottal aerodynamic pressure are presented and the stresses in the vocal fold tissue are analyzed. For simplicity, no fluid structure interaction is considered here, and the main attention is concentrated on the magnitudes of the vocal folds displacements in the longitudinal and lateral directions, and on the normal and shear stresses. The stresses are compared with the impact stress measured on the physical vocal folds model.

2 3D FE MODEL OF THE HUMAN LARYNX

The geometry and relations between the arytenoids, thyroid and cricoid cartilages was derived from CT images of an enlarged synthetic resin model of the human larynx from the collections of the Anatomical Institute of the 3rd Medical Faculty of the Charles University in Prague and on the bases of the anatomy textbook by Standring (2004). This model is a copy of the original physical model from Germany (Deutches Hygiene-Museum, Institute für biologisch-anatomische Anschauungsmaterialien, Dresden).

The 3D complex dynamic FE model of the human larynx was developed by transferring the CT image data from the DICOM format to the FE mesh. The geometrical configuration of the cross-section of the vocal fold was taken according to Hirano (1975). Three layers of vocal fold tissue are considered (Titze, 2006): epithelium, vocal ligament and muscle with different physical and material properties - see Figure 1.

The developed fully parameterized 3D FE model enables to vary the thickness and material properties of the individual layers and to take into account longitudinal pretension and adduction of the vocal folds by positioning of the arytenoids and thyroid cartilages i see Figure 2. The initial position of the model corresponds to the following geometrical parameters: the total length of the vocal folds \( L = 18 \) mm, measured from the vocal process to the anterior commissure, the prephonatory glottal gap in the middle \( 2g0 = 0.4 \) mm, and the vocal fold thickness \( T = 7.8 \) mm.

The complete FE model of the larynx developed in the FE programme system ANSYS consists of linear and quadratic 3D volume elements for the vocalis muscle and for the vocal ligament. The shell elements were used for the epithelium and tetrahedral elements for other tissues. Contact elements were used for modelling the contact between left and right vocal fold during vibrations with collisions. Implementation of the contact elements on the vocal folds surface enabled to model the impact stresses in the vocal fold tissue layers during the collision of the vocal folds. The total number of elements were about 500 000. The parametric vocal folds model, created by finite elements for ligament and muscle and shell elements for epithelium, enables to modify the tissue structure relatively easily and to simulate some pathological changes.

The nonlinear elasticity theory for large-strain deformations with linear strain-stress relationship was implemented in the computer program. The linear transversal isotropic material model was used for the vocal fold tissue, where the matrix of the elastic constants in strain-stress relations is defined as follows:
the prephonatory glottal half-gap folds self-sustained vibrations for the airflow rate (see Horáček et al., 2005) of the vocal folds during the vocal folds vibration were characterized by the open quotient $OQ=0.725$, defined as the open time of the glottis divided by the period length of the vocal folds vibrations, by the maximum glottis opening $GO=1.27$ mm and the maximum impact stress estimated by the Hertz theory $IS=1328$ Pa. The intraglottal periodic pressure $p(y,t)$ used for excitation of the 3D FE model of the vocal folds is shown in Figure 4 for two periods of the oscillation cycle. The pressure loading of the vocal fold surface was applied to all nodes in the intraglottal region, i.e. on the surface from the inferior to the superior vocal fold margin and along the length of the vocal folds.

$[\mathbf{e}] = \begin{bmatrix} E_y & -\mu_y E_y & -\mu_y E_y \\ -\mu_y E_y & E_y & -\mu_y E_y \\ -\mu_y E_y & -\mu_y E_y & E_y \end{bmatrix}$

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$\sigma_{e} = [\begin{bmatrix} \sigma_{xx} \\ \sigma_{yy} \\ \sigma_{zz} \end{bmatrix}]$

$\nu \epsilon_\eta (1+\nu_\eta) \sigma_{\eta}$

$\nu = 0.01$ $\epsilon_\eta = 0.01$

Table 1. Nominal values of material constants of individual tissue layers according to Mital (2008) - E=Epithelium, L=Ligament, M=Muscle, C=Cartilage, LT=Loose connective Tissue.

<table>
<thead>
<tr>
<th></th>
<th>E</th>
<th>L</th>
<th>M</th>
<th>C</th>
<th>LT</th>
</tr>
</thead>
<tbody>
<tr>
<td>$G_p$ [kPa]</td>
<td>530</td>
<td>0.87</td>
<td>1.05</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>$G_l$ [kPa]</td>
<td>10</td>
<td>40</td>
<td>12</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>$\epsilon_p$</td>
<td>0.3</td>
<td>0.3</td>
<td>0.3</td>
<td>0.47</td>
<td>0.4999</td>
</tr>
<tr>
<td>$F_p$ [kPa]</td>
<td>26</td>
<td>104</td>
<td>31</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>$f$ [kg/m$^3$]</td>
<td>1020</td>
<td>1020</td>
<td>1020</td>
<td>1020</td>
<td>1020</td>
</tr>
<tr>
<td>$\epsilon_{pl} = \epsilon_{pL}$</td>
<td>0.3</td>
<td>0.3</td>
<td>0.3</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

3 FREQUENCY - MODAL CHARACTERISTICS OF THE MODEL

The modal analysis and consequently also the numerical simulations were performed for the pre-stressed vocal folds. The tissue prolongation in the longitudinal $z$ direction $\bar{U} = 5\%$ of the original length $L$ of the vocal folds was realized by rotating the thyroid cartilage against the cricoid cartilage. The frequency-modal properties of the FE model are presented for the first four eigenfrequencies and the mode shapes of vibration in Figure 3. Because the model is not perfectly symmetric, the fundamental eigenfrequencies for the left and right vocal fold are slightly different, however the mode shapes are very similar. The displacements in the vertical $y$ and horizontal $x$ directions are dominant for the first and second eigenmode, respectively.

4 NUMERICAL SIMULATIONS OF VOCAL FOLDS VIBRATION

4.1 Driving intraglottal pressure

The motion of the vocal folds was numerically simulated for a prescribed intraglottal pressure given by a periodic function in the time domain. The intraglottal pressure signal loading the vocal fold surface was generated by the 2D aeroelastic model (see Horáček et al., 2005) of the vocal folds during the vocal folds self-sustained vibrations for the airflow rate $Q=0.179$ l/s, the prephonatory glottal half-gap $g_0=0.2$ mm and the fundamental frequency $F_0=100.766$ Hz, that corresponded to the subglottal pressure $P_{sub}=378.4$ Pa, and the resulted vocal folds vibration were characterized by the open quotient $OQ=0.725$, defined as the open time of the glottis divided by the period length of the vocal folds vibrations, by the maximum glottis opening $GO=1.27$ mm and the maximum impact stress estimated by the Hertz theory $IS=1328$ Pa. The intraglottal periodic pressure $p(y,t)$ used for excitation of the 3D FE model of the vocal folds is shown in Figure 4 for two periods of the oscillation cycle. The pressure loading of the vocal fold surface was applied to all nodes in the intraglottal region, i.e. on the surface from the inferior to the superior vocal fold margin and along the length of the vocal folds.

$M \ddot{u} + B \dot{u} + K u = P(x, y, z, t)$

where $M$, $B$, $K$ are the global mass, damping and stiffness matrices, $u$ is the vector of nodal displacements and $P$ is the...
vector of nodal excitation forces given by prescribed the intraglottal pressure. The damping matrix $B$ was chosen proportional to matrix $M$ and $K$:

$$B = \alpha M + \beta K,$$

(3)

where Rayleigh damping constants where $\alpha=31$ and $\beta=0.001$.

First the simulation accuracy was tested by increasing the mesh density of the vocal folds tissue and by decreasing the time step of the integration $\Delta t$.

Ten pulses were considered as sufficient in order to obtain a periodic dynamic time response of the system.

### 4.3 Computed displacements

The computed trajectories $u_x(t)$, $u_y(t)$, $u_z(t)$ in the selected nodes of the vocal fold tissue during stabilized oscillation cycles are shown in Figure 5 and the displacements of the node 1 are presented in Figure 6 in time domain.

![Figure 5. Numerically simulated plane (xy) trajectories of the right vocal fold in the middle cross-section.](image)

The computed normal stresses $\hat{\sigma}_{xx}$, $\hat{\sigma}_{yy}$ and the shear stresses $\hat{\tau}_{xy}$ are shown in time domain in Figure 7.

Because of a significant phase shift (ca 180 deg) between the vertical and horizontal motion of the vocal folds (see Figs. 5 and 6) the stresses maxima are not reached during the vocal folds contact but in the open phase of the glottis. The normal stresses $\hat{\sigma}_{xx}$ and $\hat{\sigma}_{yy}$ have the compression maxima in the ligament (node 2) of about -0.8 kPa and -0.4 kPa, respectively, and only slightly lower values are presented in the muscle (node 3), see Fig. 7. The maximum of the tensile stress $\hat{\sigma}_{xx} \cong 0.3$ kPa is in the muscle. The maximum of the shear stresses $\hat{\tau}_{xy} \cong 0.6$ kPa and $\hat{\tau}_{xy} \cong 0.5$ kPa were obtained in the ligament (node 2) and in the muscle (node 3), respectively. From the perspective of tissue damage, the most dangerous tensile and shear stresses appeared in the open phase of the glottis.

The distribution of the normal stresses $\hat{\sigma}_{xx}$, $\hat{\sigma}_{yy}$, and the shear stresses $\hat{\tau}_{xy}$ inside the vocal fold tissue in the middle cross-section is shown in the Figure 8 at the time instant of a maximal opening of the glottis and in Figure 9 at the time instant of a vocal folds position during collision.

An important cumulative stress in strength and fracture mechanics of materials is the so called equivalent stress $\hat{\sigma}$ or Von Mises stress $\hat{\sigma}$, defined as a function of all components of the stress tensor:

$$\hat{\sigma}_{eq} = \sqrt{\frac{1}{2} \left[ (\sigma_{xx} - \sigma_{yy})^2 + (\sigma_{yy} - \sigma_{zz})^2 + (\sigma_{zz} - \sigma_{xx})^2 + 6(\sigma_{xy}^2 + \sigma_{yz}^2 + \sigma_{zx}^2) \right]}.$$

(4)

The computed equivalent stresses in the ligament (node 2) and in the muscle (node 3) are shown in Figure 10 in time domain during the oscillation cycles considered. The maximum equivalent stress of about 6 kPa was obtained in the ligament (node 2) reaching this stress magnitude during the impact. However, in the muscle (node 2) the maximum $\hat{\sigma}_{eq} \cong 2.3$ kPa was obtained in open phase of the glottis oscillation period. The equivalent stress inside the tissue decreases with the depth under the vocal fold surface, see $\hat{\sigma}_{eq}$ in the nodes 2 and 3. We can also note here, that the differences between the maximum and minimum of the equivalent stresses $\hat{\sigma}_{eq} \cong 2.5$ kPa in the ligament and $\hat{\sigma}_{eq} \cong 1.1$ kPa in the muscle correspond to the dynamics of the system pre-loaded by the contact (impact) forces during the vocal folds collisions, which are superimposed on the pre-stress due to the static vocal folds pre-tension, i.e. of about 5 kPa in the ligament and 1.5 kPa in the muscle (see Fig. 10).
From the point of estimation of the material damage initialization and fatigue crack propagation, knowledge of the principal normal stresses is very important. The three principal normal stresses were calculated from the tensor stress solving the condition:

\[
\begin{vmatrix} \sigma_{xx} - \lambda & \sigma_{xy} & \sigma_{xz} \\ \sigma_{yx} & \sigma_{yy} - \lambda & \sigma_{yz} \\ \sigma_{zx} & \sigma_{zy} & \sigma_{zz} - \lambda \end{vmatrix} = 0.
\]  
\( (5) \)

The principal normal stresses \( \sigma_1(t) \), \( \sigma_2(t) \) and \( \sigma_3(t) \) computed in the vocal fold tissue are presented in time domain in Figure 11. Maximum of the most dangerous tensile principal stress \( \hat{U}_L \) \( \cong 6 \) kPa appears in the ligament (node 2) during the vocal folds collision. Maximum magnitudes of the tensile principal stress \( \hat{U}_L \) \( \cong 1.8 \) kPa in the muscle (node 3), are much less, but they appeared during the open phase of the glottis. The maximum compress principal stresses are seen in \( \hat{U}_L(t) \) during

Figure 7. Computed normal stresses \( \hat{U}_{xx}(t) \), \( \hat{U}_{yy}(t) \) and shear stresses \( \hat{U}_{xy}(t) \) in the ligament (node 2) and in the muscle (node 3).

Figure 8. Distribution of the normal stresses \( \hat{U}_{xx} \) (upper panel) and \( \hat{U}_{yy} \) (middle panel) and shear stress \( \hat{U}_{xy} \) (lower panel) computed in the middle cross-section of the vocal folds for the open glottis at the time instant \( t = 1.06613 \) s.
the open phase of glottis oscillation, up to about $\sigma_3 \approx -1$ kPa in the ligament and $\sigma_3 \approx -0.9$ kPa in the muscle.

In conclusion, none of the sixth stresses can be simply neglected; all stresses create a complicated 3D loading of the vocal fold tissue that changes during the vibration cycle, creating a complicated stress-strain wave propagating in the tissue in the anterior-posterior and inferior-superior directions, similarly like the mucosal waves on the vibrating vocal folds that may move in anterior-posterior, mediolateral and in caudal-cranial direction (Sonininen and Laukkalanen, 2003).

Figure 9. Distribution of the normal stresses $\sigma_{xx}$ (upper panel) and $\sigma_{yy}$ (middle panel) and shear stress $\sigma_{xy}$ (lower panel) computed in the middle cross-section of the vocal folds for the closed glottis at the time instant $t = 1.06987$ s.

Figure 10. Computed equivalent stresses $\sigma_{eqv}$, in the ligament (node 2) and in the muscle (node 3).

5 EXPERIMENTAL SIMULATIONS OF VOCAL FOLDS VIBRATION

The in vitro measurements of voicing were performed on a developed artificial larynx based on the CT images of human larynx taken during phonation (Guzman et al. 2013). The CT examination was done for a trained male singer (34 years old). The subject was placed in a CT scanner in supine position and phonated a sustained vowel [a:] in a habitual pitch and comfortable loudness. The CT images were segmented and processed into volume model of the vocal tract (Vampola et al. 2013).

The 1:1 scaled model of the larynx was prepared of the commercial addition (platinum)-cure two-component silicon rubber Ecoflex 00-10 (Smooth-On, US). The mixture was casted into the mould made by 3D printing from the volume model of the human vocal tract. After mixing, the viscous liquid was poured into the tube-like mould and degassed in vacuum oven to get rid of the bubbles and then sealed with wax and cured for 4 hours at the room and at the last 30 minutes at the temperature elevated to 50°C. After curing and cutting out the part above the vocal folds with a sharp puncher, the rubber model was ready for tests (see Figures 12 and 13).

The viscoelastic properties of the silicone rubber were measured by oscillatory rheometer. The complex shear modulus $G = G' + i G''$ close to the phonation frequency 100 Hz revealed significant viscous response, at $G' = 11$ kPa and $G'' = 3.5$ kPa.

The experiments with the artificial larynx were performed in a test stand that enables synchronous registration of the airflow induced vocal fold vibrations using a high speed camera, measurement of impact stress between the vocal folds during collisions, the subglottal dynamic and mean air pressure and the generated acoustic signal (Horáček et al. 2013).
Figure 11. Principal normal stresses $\sigma_1(t)$, $\sigma_2(t)$ and $\sigma_3(t)$ computed in the ligament (node 2) and in the muscle (node 3).

Figure 12. a) Mould produced by 3D-printing and b) the casted larynx model made of silicone rubber.

Example of the simultaneously recorded time signals for the impact stress ($IS$) measured by a miniature pressure transducer, the subglottal pressure ($P_{sub}$) and the glottis opening ($GO$) evaluated from the images of the self-oscillating vocal folds are presented in Figure 14. When the glottis was opened during the vibration period, the contact sensor measured the intraglottal air pressure of about 2.5 kPa in the airflow between the vocal folds. As a result the maximum of impact stress was $Max IS \cong 2.6$ kPa. The maximum of $P_{sub} \cong 1.7$ kPa was delayed after the $Max IS$ and the maximum $GO \cong 1.5$ mm was delayed behind $P_{sub}$.

Depending on the mean flow rate $Q = 0.15-0.65$ l/s the measured $Max IS \cong 0-7.5$ kPa and the sound pressure level $SPL \cong 60-92$ dB of the acoustic signal in a distance of 20 cm from the vocal folds were increasing approximately linearly. The mean subglottal pressure varied in the interval $P_{sub} \cong 1-1.83$ kPa and the fundamental frequency of the vocal folds self-oscillations varied in the interval $F_0 \cong 94-98$ Hz.

The measured phonation characteristics are in good agreement with the values found in human excised larynges (Reed et al. 1992; Verdolini et al. 1999).

Figure 13. Larynx replica with self-oscillating vocal folds at the time instants of minimum and maximum glottis opening.
The computed normal and shear stresses show a significant triaxial state of stress in the vocal fold tissue and no compound of the stress tensor can be neglected. Three triaxial state of stress in the vocal fold tissue and no

6 CONCLUSIONS

The geometry of the parametric 3D FE model of the vocal folds developed as a part of the complex larynx model can be easily modified, enabling tuning and optimisation procedures for finding proper model geometric and material parameters related to the vocal fold vibration characteristics.

The results suggest that the model enables to predict stresses in the layered vocal fold tissue due to the vibration of the vocal folds in phonation regimes with collisions. The motion of the vocal folds excited by periodic intraglottal pressure pulses seems to be qualitatively similar to the vibration patterns known from clinical observations.

The computed normal and shear stresses show a significant triaxial state of stress in the vocal fold tissue and no compound of the stress tensor can be neglected. Three principal normal stresses are periodically changing their magnitudes and direction. The maximum stresses were found mainly in the ligament near the vocal folds surface, decreasing with the depth inside the tissue. The important finding is that the maximum stresses are not present only in the closed phase of the vocal folds motion during their collision but approximately similar maximum magnitudes of the stresses were found in the opening phase of the glottis.

The changes of the stresses due to the vocal folds collisions were found in a good general agreement with the impact stress measured in the rubber vocal folds model. The differences 2.5 kPa found between the computed contact and contactless equivalent stresses (recall Fig. 10) were close to the maximum impact stress 2.6 kPa measured in the artificial larynx.

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