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Mychal Spencer

Thomas Siegmund

Purdue University, siegmund@purdue.edu

Luc Mongeau

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Determination of superior surface strains and stresses, and vocal fold contact pressure in a synthetic larynx model using digital image correlation

Mychal Spencer, Thomas Siegmund,^{a)} and Luc Mongeau^{b)}

School of Mechanical Engineering, Purdue University, West Lafayette, Indiana 47907

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Stresses and strains within the vocal fold tissue may play a critical role in voice fatigue, in tissue damage and resulting voice disorders, and in tissue healing. In this study, experiments were performed to determine mechanical fields on the superior surface of a self-oscillating physical model of the human vocal folds using a three-dimensional digital image correlation method. Digital images obtained using a high-speed camera together with a mirror system were used to measure displacement fields, from which strains, strain rates, and stresses on the superior surface of the model vocal folds were computed. The dependence of these variables on flow rate was established. A Hertzian impact model was used to estimate the contact pressure on the medial surface from superior surface strains. A tensile stress dominated state was observed on the superior surface, including during collision between the model folds. Collision between the model vocal folds limits the medial-lateral stress levels on the superior surface, in conjunction with compressive stress or contact pressure on the medial surface. © 2008 Acoustical Society of America.

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I. INTRODUCTION

Voice is produced by the dynamic interactions between lung-driven airflow, sound waves in the airway, and the deformations of the vocal folds tissue (VFs). A good understanding of the basic mechanics of phonation is needed for the development of better clinical diagnosis tools and voice prosthetic devices. Accurate measures of the mechanical stress in VFs are essential for the understanding of vocal fold damage and healing (Titze, 1994).

The present study is concerned with the deformation and stresses in a self-oscillating physical model of vocal folds. In general, the mechanical stresses within the VFs depend on: (1) the fluid pressure within the glottis; (2) the occurrence of impact between the vocal folds, and the associated change in linear momentum of the bulk of the vocal folds; and (3) elastic stresses related to the deformation of the vocal fold tissue due to posturing. These factors have been investigated in a number of previous studies. Studies on airflow through the glottis have been the focus of many investigations related to human voice production, mainly because airflow is a primary factor in the generation of sound (Titze, 2000). Flow studies are relevant for the present investigation since they provide first-order insight into the pressure applied onto the vocal fold. For modeling purposes, the glottis may be construed as an orifice with a time-varying geometry. The flow through such orifices is complex, involving jet formation, turbulence, dynamic flow separation, and asymmetry. Time varying glottal jet plume angles may be due to asymmetric

orifice contours or the so-called Coanda effect. The sound pressure produced during voicing is a function of the time varying changes in the major loss coefficient of the glottal orifice, as described by Park and Mongeau (2007). Previous research using physical laryngeal models, rigid models, and computational models has yielded a good understanding of the fluid pressure distribution acting over the surface of the vocal folds (Scherer *et al.*, 2001). Typically, the fluid pressure on the inferior surface is near the subglottal pressures, and oscillates around the mean transglottal pressure (typically 0.8 kPa above atmosphere for normal voice). The fluid pressure on the superior surface oscillates, with peak values around 0.3 kPa, and a mean value slightly (around 0.1 kPa) above the ambient atmospheric pressure. The pressure distribution along the medial surface has been well documented, and involves a gradual decrease in pressure that is greater than the transglottal pressure (by 0.2 to 0.3 kPa) due to convective flow acceleration within the orifice (i.e. the so-called “Bernoulli effect”), followed by a fairly abrupt pressure recovery downstream of the region of flow separation. These flow studies have furthermore demonstrated that the geometry of the vocal fold model considered would indeed be relevant. Providing a step toward a more realistic and standardized modeling framework, Scherer *et al.* (2001) proposed a vocal fold geometry, referred to as M5, and investigated fluid flow through rigid static models. In the M5 model the vocal folds are represented as prisms with trapezoidal base. Fulcher *et al.* (2006) studied the response of spring mass models with excitation pressures based on measured M5 wall pressure data. A deformable continuum vocal fold model adopting the same basic geometry was proposed by Thomson *et al.* (2005). There experiments were conducted using deformable, self-oscillating models to study the energy

^{a)}Electronic mail: siegmund@purdue.edu

^{b)}Present address: Department of Mechanical Engineering, McGill University, Montreal, Quebec H3A 2K6, Canada.

transfer mechanisms between the flow and the radiated sound, and to verify detailed computer models of the coupled fluid–structure interactions. It was demonstrated that the cyclic deformation of the deformable vocal fold model is essential in the energy transfer from flow to solid, and the subsequent development of stresses and strain in the vocal fold. To investigate the details of the VF stress distribution due to flow induced deformation, numerical studies of fluid–structure interactions within continuum laryngeal models have been performed. Of particular interest to the present study are investigations related to vocal fold contact. The first work aiming to determine vocal fold contact pressure is that of [Jiang and Titze \(1994\)](#). This work outlined the three phases in vocal fold contact—an impact phase, a progressive pressure buildup phase, and an open phase. In [de Oliveira Rosa et al. \(2003\)](#) VF contact was considered, and the importance of incomplete closure was highlighted but no details on the resulting deformation fields was given. Details of dynamic contact conditions in individual vibration cycles—neglecting fluid pressure—were investigated by [Gunter \(2003\)](#) using finite element models with high spatial resolution. Alternatively, [Horáček et al. \(2005\)](#) employed a lumped mass mechanical representations of the VFs based on rigid mass–spring–damper systems ([Flanagan and Landgraf, 1968](#); [Ishizaka and Flanagan, 1972](#)) with contoured rigid walls to investigate vocal fold collision conditions and the associated normal tractions on the medial surfaces of the vocal folds during collision. These authors predicted impact pressure values of the order of 3 kPa. The treatment of the vocal folds as rigid bodies, however, does not allow the effects of the deformation of the vocal folds to be considered. Later, a refined analysis of impact stress was performed by [Tao et al. \(2006\)](#) using a dynamic finite element model that included contact and coupled flow–structure interactions. Contact pressures reported in that study range up to 6 kPa. Direct experimental measurements of the contact pressure have been reported ([Verdolini et al., 1999](#)) using a miniature pressure transducer *in vivo*. Measured contact pressure values along the membranous portion of the vocal folds ranged from 0.5 to 3.0 kPa. Subsequently, [Jiang et al. \(2001\)](#) related pressure sensor data to photoglottography.

Imaging techniques for the measurement of VF deformations are of interest for possible clinical protocols. Progress has been made in the use of optical techniques for the study of VF vibrations using (endoscopic) high-speed imaging ([Tigges et al., 1996](#)). This method demonstrated that vocal fold motion, notably the movement of the free vocal cord edge, can be tracked during the oscillatory cycle. Videokymography has been used to construct digital image sequences for the visualization of vibration patterns. [Švec et al. \(2000\)](#) demonstrated the usefulness of this method by qualitatively relating vibrational modes to the magnitude of vocal fold deformation. Attempts have been made to use laser-based optorelectrometry to determine VF displacement ([Ouaknine et al., 2003](#)). This method yielded qualitative information on the inferior–superior surface motion. Methods based on triangulation algorithms, and the tracking of a single laser-generated speckle and its reflected image have allowed the measurements of the displacement of individual

points on the superior surface of the VFs ([Manneberg et al., 2001](#)). Full-field measurements of tissue deformation have been made using a microsuture technique on exercised canine larynges; the results were used to study the different vibrational modes of the VF medial surface ([Berry et al., 2001](#)) and provided insight into mucosal wave propagation. The use of the electronic speckle pattern interferometry to analyze VFs has been proposed ([Gardner et al., 1995](#)). This method has been found suitable, in principle, for noninvasive, full field strain measurements, but is limited to small deformation magnitudes.

These limitations can be overcome by the use of digital image correlation. Digital image correlation is a method that provides full field displacement fields on the surface of deformable structures. The method is noncontact and required that a speckle pattern be deposited on the structure of interest, e.g., [Sutton et al. \(1986\)](#). [Mantha et al. \(2005a, b\)](#) first used this method to study conditions on the superior surface of the vocal fold model of [Thomson et al. \(2005\)](#). [Berry et al. \(2006\)](#) used a similar VF model to study medial surface dynamics, also using a digital image correlation (DIC) method.

The goal of the present study was to conduct full field, time resolved measurements of the structural deformation of a VF model during self-oscillation using digital image correlation. This method is advantageous since it allows for large displacement magnitudes. Digital images were obtained using a high speed digital camera. Three-dimensional (3D) DIC was performed by projecting image pairs onto a single image frame using a mirror system. Images of the superior surface of a VF physical model were considered, because this surface is visible in laryngoscopic evaluations. While the displacements are the primary measured quantities in DIC, strains and strain rates can be obtained by appropriate data processing.

A deformable physical model of the human VFs was used in order to study the mechanical response in a controlled setting, without the need for animal models or excised larynges. The material used to produce the vocal fold model was silicone rubber. Silicone models are convenient for fundamental investigations, at the expense of a lack of realism in the tissue properties and the vocal fold geometry. The model vocal folds were embedded into rigid Plexiglas™ supports. The test rig supporting the laryngeal model did not allow tension adjustments of the vocal folds. This model was nevertheless deemed appropriate for the main goal of the investigation, which was to determine the usefulness of 3D DIC for the determination of mechanical states of the superior surface of the model, and to establish a process to obtain contact pressure data from superior surface data.

II. METHODS

A. The physical vocal folds model

The M5 vocal fold geometry reported by [Scherer et al. \(2001\)](#) was adopted for construction of the synthetic vocal folds. Since several other investigators have employed this model, [Thomson et al. \(2005\)](#), [Fulcher et al. \(2006\)](#), it allows for comparison between independent studies. Each fold con-

(a)

(b)

(c)

FIG. 1. (a) Characteristic dimensions of the model vocal folds, (b) experimental setup, and (c) mirror system.

sists of a prismatic extrusion of an approximately trapezoidal base with the inclined side facing the airflow, as illustrated in Fig. 1(a). The coordinate system is such that the x , y , and z axes point along the medial-lateral, the anterior-posterior, and the inferior-superior direction (i.e., the flow direction), respectively. The model inferior-superior length was $t = 10.7$ mm, its medial-lateral length was $w = 8.4$ mm, and the VF length was $L = 22$ mm. This length of the vocal fold is on the upper bound of physiological data, [Zhang et al. \(2006\)](#). However, while the model possesses rigid boundaries an actual vocal fold does not. The anisotropic multilayer VF structure was simplified using an isotropic and homogeneous solid. The model VFs were cast from silicone rubber (Ecoflex; Smooth-On, Inc.) with a 1:4 ratio of liquid silicone rubber to silicone thinner, selected to obtain material properties with a modulus similar to that of soft tissue. The silicone rubber was characterized as an incompressible elastic solid material ([Kerdok et al., 2003](#)). The shear modulus G was determined through instrumented hardness testing (ELF 3200; Bose) with a flat cylindrical indenter of radius r ([Pawlak and Keller, 2002](#)). The shear modulus was obtained from

$$G = \frac{1}{8r} \left(\frac{dP}{d\delta} \right), \quad (1)$$

where the indentation force is P , the indentation depth is δ , and $dP/d\delta$ is the slope of a P - δ regression. The indentation tests were performed for the conditions $\delta_{\max} = 0.85$ mm, $\dot{\delta} = 0.15$ mm/s, and $r = 3.0$ mm. The shear modulus was found to be $G = 2.0$ kPa, based on five repeated experiments. When incompressibility is assumed, these measured properties are consistent with an elastic modulus of $E = 6.0$ kPa. The model VFs were embedded into Plexiglas enclosures of thickness t with rectangular cut-outs of dimensions $L \times w$. Two symmetric enclosures were clamped using screws running in the medial-lateral direction within the plates. The outward normal of the VF superior surface coincided with the flow direction. In absence of flow, there was no visible gap between the VFs in the prephonatory state. Talc was applied to the medial surface to minimize the amount of adhesion during collision.

B. Flow supply and optical measurement setup

Airflow was generated by a centrifugal compressor. A 36-cm-diam and 48-cm-length cylindrical plenum chamber lined with 2.54-cm-thick fiberglass was used as an expansion chamber muffler to attenuate pressure fluctuations associated with noise in the flow supply. The flow path downstream of the muffler consisted of a 15-m-long flexible tube, and a rigid rectangular test section of cross section 23×22 mm and length 200 mm, to which the Plexiglas enclosure containing the model VFs was attached. Key components of the experimental setup are shown in Fig. 1(b). The tube termination was nearly anechoic. The measured reflection factor was approximately 0.15 at the self-oscillation frequencies observed in the experiment. Analog pressure regulators downstream of the compressor were used to control the airflow. A volumetric flow meter (MKS 558A; MKS Instruments) located upstream of the expansion chamber was used to measure the flow rate. Flush-mounted microphones were located 3.2 and 12 cm from model VFs upstream. The output signals from the two microphones, the volumetric flow meter, and a frequency counter (Fluke 45 DDM; John Fluke Mfg. Co., Inc.) were fed to a data acquisition system (Siglab 20-22A; Spectral Dynamics Inc.) sampling at 10 000 Hz with a voltage resolution of 0.01 V. For verification experiments, a laser Doppler velocimetry system (OFV 3000; Polytec Inc.) was employed.

Images of the superior surface of the model VF were obtained using a CMOS high-speed digital camera (Memrecam FX K3; NAC Image Technology Inc.) recording at 3000 frames/second with an open shutter. Images obtained with the high speed camera were processed to provide kymograms. Thereby, a single pixel row was obtained from each image and subsequent pixel rows were stacked on each other. The pixel size of the image chip was 0.011×0.011 mm. The pixel size relates to the resolution of the displacement field and the speckle size [the resolution of the DIC method is sensitive to the speckle size, e.g., see [Hung and Voloshin \(2003\)](#)]. A lens with focal length of 105 mm

(AF Micro Nikkor; Nikon Inc.) was used. A two-channel fiber optic goose-neck light with a quartz source (Model 21AC; Edmund Optics Inc.) provided uniform illumination. The orientation of the lights was optimized to reduce glare. Due to the remote location of the quartz light source a fiber optic light system avoids heating of the illuminated surface. Image pairs of the VF superior surface in undeformed and deformed configurations were obtained. A mirror system, composed of a central right angle prism with mirror coatings on the two legs and two lateral mirrors, was placed in the optical path, Fig. 1(c). Images from two viewing directions (from the left and the right side) were thus projected onto the single CMOS array.

The stereo DIC system based on a single camera with a mirror system requires a particular calibration procedure. The intrinsic parameters (i.e., focal length, aspect ratio, image center, and lens distortion) were calculated by using a grid target (a square grid of uniformly spaced white dots on a black background) positioned at various angles relative to the optical axis of the camera lens without the mirror system in place. The extrinsic parameters (i.e., the relationship between the two images produced by the mirror system) were then obtained by using a target with random speckle patterns, and two hash marks separated by a known distance, with the mirror system in place. Further details are documented in [Spencer \(2007\)](#).

The model VF superior surfaces were prepared with a random black/white speckle pattern. The silicone rubber was colored with a white pigment (Silc Pig; Smooth-On, Inc.). A spray of black enamel paint (1249 Flat Black Spray Enamel Paint; Testors) was deposited onto the superior surface resulting in a distribution with speckle diameter approximately 0.088 mm, corresponding to around 7 to 8 pixels.

C. Digital image correlation and analysis

The 3D-DIC analysis was performed using the commercially available program VIC-3D (Correlated Solutions Inc.). 3D DIC is a method that allows for measurements of the deformation of the surface of an object. Information regarding the algorithms used in this program can be readily found in the literature, i.e., in [Chu et al. \(1985\)](#), [Sutton et al. \(1986\)](#), [Helm et al. \(1996\)](#), on the in-plane correlation and in [Schreier et al. \(2004\)](#) on the stereoscopic aspects. A subset size, i.e., the size of the interrogation window in the numerical algorithm, was selected to be 23×23 pixels. All variables were obtained at a distance in excess of 0.5 mm away from the medial surface.

The output of the program includes the Lagrangian displacement field $\underline{u}(t) = (u, v, w)$ at each time step, obtained from $\underline{u} = \underline{x} - \underline{x}_0$ where $\underline{x}_0 = (x_0, y_0, z_0)$ and $\underline{x}(t) = (x, y, z)$ represent global undeformed and deformed configurations, respectively. The velocity of points on the superior surface, $\dot{\underline{u}} = (\dot{u}, \dot{v}, \dot{w})$, was obtained numerically using a fourth order finite difference method. Strains ($\varepsilon_{xx}, \varepsilon_{yy}, \gamma_{xy}$) on the VF superior surface were calculated using ([Sutton et al., 2001](#))

$$\varepsilon_{xx} = \frac{\partial u}{\partial x} + \frac{1}{2} \left\{ \left(\frac{\partial u}{\partial x} \right)^2 + \left(\frac{\partial v}{\partial x} \right)^2 + \left(\frac{\partial w}{\partial x} \right)^2 \right\},$$

$$\varepsilon_{yy} = \frac{\partial v}{\partial y} + \frac{1}{2} \left\{ \left(\frac{\partial u}{\partial y} \right)^2 + \left(\frac{\partial v}{\partial y} \right)^2 + \left(\frac{\partial w}{\partial y} \right)^2 \right\},$$

$$\gamma_{xy} = \frac{1}{2} \left\{ \left(\frac{\partial u}{\partial y} + \frac{\partial v}{\partial x} \right) + \left(\frac{\partial u}{\partial x} \frac{\partial u}{\partial y} + \frac{\partial v}{\partial x} \frac{\partial v}{\partial y} + \frac{\partial w}{\partial x} \frac{\partial w}{\partial y} \right) \right\}. \quad (2)$$

The incompressibility condition, $\varepsilon_{xx} + \varepsilon_{yy} + \varepsilon_{zz} = 0$, allowed for the determination of the third strain component, $\varepsilon_{zz} = -\varepsilon_{xx} - \varepsilon_{yy}$, which yielded the complete strain vector $\underline{\varepsilon} = [\varepsilon_{xx}, \varepsilon_{yy}, \varepsilon_{zz}, \gamma_{xy}, 0, 0]^T$. Strain rates were again obtained from the strain time histories using a fourth-order finite difference method.

For an incompressible material, the deviatoric stress, $\underline{\sigma}^d = (\sigma_{xx}^d, \sigma_{yy}^d, \sigma_{zz}^d, \tau_{xy}, 0, 0)^T$, and the hydrostatic stress, σ^h , need to be determined independently [see, for example, [Zienkiewicz and Taylor \(2000\)](#)], in order to obtain the stress state $\underline{\sigma} = (\sigma_{xx}, \sigma_{yy}, \sigma_{zz}, \tau_{xy}, 0, 0)^T$. The stress state for the superior surface can be obtained by assuming the pressure on the superior surface, p_{supra} , to be known and identified with the stress component $\sigma_{zz} = p_{\text{supra}}$. Previous studies ([Scherer et al., 2001](#)) of steady flows through glottis-shaped orifices have shown that p_{supra} is approximately equal to the atmospheric pressure, $p_{\text{supra}} = \sigma_{zz} \approx 0$ on the superior surface. When there is no supraglottal acoustic load (i.e., if there is no confining air duct at the orifice discharge) the radiated sound pressure is also insignificant due to the poor radiation efficiency of the pulsed jet ([Mongeau et al., 1997](#)). The hydrostatic stress σ^h was thus approximated as $\sigma^h = \sigma_{zz} - \sigma_{zz}^d = p_{\text{supra}} - \sigma_{zz}^d \approx -\sigma_{zz}^d$. The remaining stress components $\sigma_{xx}, \sigma_{yy}, \tau_{xy}$ were obtained from $\underline{\sigma} = \underline{\sigma}^d + \sigma^h \underline{m}$, with the deviatoric stress $\underline{\sigma}^d$ equal to

$$\underline{\sigma}^d = 2G \left(\underline{I}_0 - \frac{1}{3} \underline{m} \underline{m}^T \right) \underline{\varepsilon}. \quad (3)$$

The 6×6 matrix $\underline{I}_0$ possesses entries only along its diagonal, with the first three entries equal to one and the remainder equal to $1/2$, and $\underline{m} = [1, 1, 1, 0, 0, 0]^T$. As shown in the Appendix, the superior surface stresses are given by

$$\frac{\sigma_{xx}}{G} \approx 4\varepsilon_{xx} + 2\varepsilon_{yy} \quad (4)$$

and

$$\frac{\sigma_{yy}}{G} \approx 4\varepsilon_{yy} + 2\varepsilon_{xx}. \quad (5)$$

D. Evaluation of measurement accuracy

Errors in high speed DIC measurements were investigated by [Siebert et al. \(2007\)](#), and may be categorized as (1) correlation errors and (2) reconstruction errors. Correlation errors include background noise in the image (which can be enhanced by increasing the image gain) and the quantization of grayscale values. In [Siebert et al. \(2007\)](#), either one of these errors was less than 0.05 pixels, at worst. The reconstruction errors were caused by uncertainties in the calibration of the 3D-DIC system. It was shown that reconstruction errors were on the order of 0.02 pixels. It is assumed that errors were of the same magnitude and deemed to be negli-

FIG. 2. Inferior-superior velocity from LDV (---) and numerical differentiation of DIC displacement data (—). The insert photograph shows the vocal fold model with the measurement locations for LDV. Results are for point (1). Instants of open (A), closed (B), and maximum surface velocity (C) are indicated.

gible in the present study, despite the differences in optics and equipment. Comparisons between results from DIC and a strain gauge for a tensile test indicated maximum strain differences of 0.03 millistrains. The influence of speckle size on the accuracy of DIC was investigated by Hung and Voloshin (2003). It was estimated that a speckle size of roughly two to eight times the pixel size typically yields the most accurate results.

A LDV with a sampling rate of 11 800 Hz and resolution of 0.01 mm/s was used to verify the accuracy of the inferior-superior component of velocity obtained from the 3D-DIC method. Reflective patches were attached to the superior surface of a second VF model with $G=2.9$ kPa (slightly stiffer than the model described in Sec. II A), as shown in the inset photo at the upper right portion of Fig. 2. The transverse Eulerian velocity component along the direction of the laser beam, i.e., the inferior-superior direction, was measured and compared to that obtained from 3D-DIC measurements. Velocities from DIC are in the Lagrange framework (material point tracking) while LDV data are in the Eulerian framework and track a fixed region in space. In order to reduce the noise associated with the finite difference approximation in the DIC data, a low-pass filter (127 point linear-phase FIR filter using the Parks-McClellan algorithm with passband frequency of 200 Hz and stopband frequency of 400 Hz) was incorporated. Overall good agreement between the velocities from the LDV and the 3D-DIC analysis was obtained, as shown in Fig. 2. RMS values of 0.1326 and 0.1469 were found for the filtered DIC and LDV data, respectively. This implies a relative error of 9.7%. The errors were attributed to the influence of the addition of the reflecting patches, which seemed to affect the VF vibration, and the fact that LDV averages velocity over the beam diameter. The change in the orientation of the superior surface during vibration, and the errors introduced by the fourth-order finite

FIG. 3. Kymograms at: (a) $Q=406$ cm³/s (the phonation onset), and (b) $Q=690$ cm³/s. Supplementary images indicate kymogram location: $z = \text{const}$ at $y=0$. Distance $2a$ indicates the anterior-posterior extension of the contact area.

difference scheme may also have contributed. It was felt that the errors were mostly intrinsic to the experimental design, and not representative of the accuracy of the experimental data.

III. RESULTS

A. Flow, pressure, frequency, and high-speed imaging

The volumetric flow rate, Q , was increased stepwise until self-oscillation was initiated, as detected through the use of a frequency counterfed with the output signal from one microphone located in the subglottal region. At oscillation onset, the volumetric flow rate was $Q_{\text{th}}=406$ cm³/s, and the frequency was $f_{\text{th}}=88.9$ Hz. The corresponding onset pressure was $p_{\text{th}}=0.87$ kPa. For $Q_{\text{th}} \leq Q \leq 813$ cm³/s, the relationship between pressure p and flow rate Q was characterized by the linear regression $p[\text{kPa}]=0.0009Q[\text{cm}^3/\text{s}]+0.5263$, with $R^2=0.99$. High speed imaging indicated model VF collision from a flow rate, referred to as the collision onset, of $Q=Q_C=550$ cm³/s. The oscillation frequency reached a maximum, $\max(f)=90.1$ Hz, at $Q=Q_C$. The frequency, f , dropped very slightly with flow rate beyond the collision onset, from 90.1 Hz to $f=88.9$ Hz at a flow rate of $Q=750$ cm³/s. The energy dissipated by collision and adhesion is most likely responsible for this trend.

The repeatability and variability of the measurements were investigated. In onset experiments, repeated flow rate increases followed by 10-min-long interruptions yielded consistent Q_{th} values. Steady, constant flow rates were maintained for periods of 45 min ($Q=\text{const} > Q_{\text{th}}$). For $Q < Q_C$ the frequency remained constant over time with a maximum deviation less than 0.1 Hz ($\Delta f < 0.1$ Hz), while for $Q > Q_C$ the frequency decreased slightly with time with a maximum deviation of 0.8 Hz ($\Delta f = -0.8$ Hz).

Kymogram images of the midline of the model ($y=0$) are shown in Fig. 3. The kymogram was constructed from one single vibration cycle; two images of this cycle—for the open and closed state—are shown in Fig. 3. The locations of the pixel lines used for the kymogram images are indicated

in Fig. 3. The kymograms clearly show differences between the vibration patterns for $Q=Q_{th}$ and for $Q=690\text{ cm}^3/\text{s} > Q_C$. At $Q=Q_{th}$, Fig. 3(a), no collision between the model VFs was observed. At the higher flow rate, Fig. 3(b), the maximum glottal opening was much increased. In the closed state, collision of the medial surfaces was clearly visible; however, closure of the glottal opening was not complete. Two distinct orifice openings at the anterior and posterior ends of the model larynx were visible during the closing stage. Kymograms, however, do not provide information on motion in the inferior–superior direction.

B. Displacements, strains, and stresses

The DIC method was used to establish the time history of the superior surface displacements in order to characterize the glottal opening process. Figure 4 shows the displacements in the inferior–superior direction, w , and the medial–lateral direction, u , for flow rates $Q=Q_{th}=406\text{ cm}^3/\text{s}$ and $Q=690\text{ cm}^3/\text{s} > Q_C$. The displacement data are for locations along a line parallel to the medial surface, 1.5 mm away from the centerline of the undeformed model VFs. Figures 4(a) and 4(b) show that, for $Q=406\text{ cm}^3/\text{s}$, the largest inferior–superior vibratory displacements occurred in the center of the model fold and VF motion, over a span of around $L/2$. The anterior and posterior extremities of the folds were almost fixed during oscillation. Figures 4(c) and 4(d) show the displacements in the medial–lateral direction. The model larynx remained in an open state over the entire cycle. The main transverse vibratory displacements occurred again in the central section of the vocal fold, over a span (of around $3L/4$) larger than that for the inferior–superior displacement. The displacements along a line parallel to the superior edge clearly show a third-order mode of vibration as visible by the three max–min values of displacement, Figs. 4(e) and 4(f).

Figures 4(e) and 4(f) show that, for $Q=690\text{ cm}^3/\text{s}$, the largest inferior–superior vibratory displacements during opening again occurred in the center of the model fold. VF motion spanned a region roughly one-half the VF length during opening. During closing, a larger portion of the VF span was actively vibrating. During closing it was observed that the curvature was reversed. Figures 4(g) and 4(h) show the displacements in the medial–lateral direction. The active span was similar to that for the onset flow rate.

Time histories of the medial–lateral and inferior–superior displacements and velocities at the midpoint of the model VF, $\underline{x}_0=(0.5,0,0)\text{ mm}$, over an oscillation period are shown in Fig. 5 for flow rates $Q=Q_{th}=406\text{ cm}^3/\text{s}$ and $Q=690\text{ cm}^3/\text{s} > Q_C$. Figure 5(a) shows the inferior–superior displacement component. The maximum displacement for both flow rates was reached at the instant of maximum open area, as for $y=0$ in Fig. 4. The deformation magnitude for $Q=690\text{ cm}^3/\text{s}$ was approximately 50% larger than that for the lower flow rate. The inferior–superior velocity component is shown in Fig. 5(b). Maximum velocities were approximately 700% larger for $Q=690\text{ cm}^3/\text{s}$ than for the lower flow rate. Figure 5(c) shows the medial–lateral displacement. Positive values generally indicate collision for

the higher flow rate. At $Q=690\text{ cm}^3/\text{s}$, the peak medial–lateral deformation was increased by 150% compared to Q_{th} . In the closed state, the positive value for $Q=690\text{ cm}^3/\text{s}$ indicates that the medial motion of the midpoint crossed the midline reference location. Figure 5(d) shows the medial–lateral velocity component. Similar to the inferior–superior velocity component, the maximum surface velocity was much increased for the higher flow rate.

The dependence of the anterior–posterior ϵ_{yy} and medial–lateral ϵ_{xx} strains on flow rate was determined for the midpoint of the model and for $Q > Q_{th}$. Midpoint values at the times corresponding to maximum (open state) and minimum (closed state) glottal areas are shown in Fig. 6. At the phonation onset $Q_{th}=406\text{ cm}^3/\text{s}$, the absolute values of ϵ_{yy}^{open} and ϵ_{yy}^{closed} were similar. For flow rates between the phonation onset and the collision onset $Q_{th} < Q < Q_C$, magnitudes of ϵ_{yy} depend linearly on flow rate. Values of ϵ_{yy}^{open} increase monotonically with Q while ϵ_{yy}^{closed} decreased with Q but remained tensile. For $Q \geq Q_C$, ϵ_{yy}^{closed} finally was compressive and reached a nearly constant value as flow rate was increased.

Midpoint values of the medial–lateral strain ϵ_{xx} in the open and closed states are shown in Fig. 6(b). These quantities were nearly zero at $Q=Q_{th}$. Values of ϵ_{xx}^{open} were compressive and increased monotonically in magnitude with flow rate. For $Q < Q_C$, values of ϵ_{xx}^{closed} were positive and increased linearly, with $|\epsilon_{xx}^{closed}| = |\epsilon_{xx}^{open}|$. For larger flow rates, above $Q=Q_C$, the values of ϵ_{xx}^{closed} reached a maximum around a value of 0.06, and decreased slightly. Extrapolation of the curve at closed state yields a medial–lateral strain estimate for vibration in absence of collision.

Strains at the midpoint of the model VF over one oscillation period are shown in Figs. 7(a) and 7(b) at $Q=Q_{th}=406\text{ cm}^3/\text{s}$ and $Q=690\text{ cm}^3/\text{s}$. The medial–lateral strain component is shown versus time in Fig. 7(a). For $Q=690\text{ cm}^3/\text{s}$, strain was constant over the time period where the orifice was closed (at least over the midspan region). The impact duration was estimated to be 0.03 s based on the period of constant medial–lateral strain during closure. Figure 7(b) shows the anterior–posterior strain component. Strains at the higher flow rate were much increased. Impact of the vocal folds is indicated by negative values. The duration of impact was consistent with that estimated from the medial–lateral strain history.

In the characterization of viscoelastic properties of soft materials strain rates are essential. Viscosity may generally be defined as the ratio of stress and strain rate. Strain rates obtained from a fourth-order finite difference approximation are shown in Figs. 7(c) and 7(d). For comparison purposes, mean values of strain rates can be estimated from the peak strain values of Fig. 6, and frequency, as $\dot{\epsilon} = 2f|\epsilon^{open} - \epsilon^{closed}|$. This approximation is useful when measurement data are limited to states where surface velocities are small (for example, due to camera shutter speed limitations). This provided reasonable estimates of the time averaged average strain rates. For example, at a flow rate of $Q=690\text{ cm}^3/\text{s}$, the mean strain rates were found to be $\dot{\epsilon}_{yy}=58\text{ s}^{-1}$ and $\dot{\epsilon}_{xx}=35\text{ s}^{-1}$, whereas the approximation from discrete values yields $\dot{\epsilon}_{yy}=62\text{ s}^{-1}$ and $\dot{\epsilon}_{xx}=33\text{ s}^{-1}$, Fig. 7. Consequently,

FIG. 4. (Color online) Inferior-superior (a), (b) and medial-lateral (c), (d) component of displacements along a line parallel and removed 1.5 mm from the medial surface at two flow rates: (1) Phonation onset $Q = 406 \text{ cm}^3/\text{s}$ and (2) $Q = 690 \text{ cm}^3/\text{s}$. The time increment between measurements is 0.363 ms.

mean strain rate values can be assessed from Fig. 6. Mean values of strain rates $\dot{\epsilon}_{yy}$ increase monotonically with flow rate. The medial-lateral mean strain rate $\dot{\epsilon}_{xx}$ reaches a limit since VF collision limits strain rate magnitudes in the medial-lateral direction. Mean strain rate values were much smaller than the peak instantaneous values, indicating a large

crest factor. This may be significant since, for Newtonian materials, strain rates are often approximated as the product of the mean strain and frequency. This, along with the fact that vocal fold tissue is non-Newtonian, could lead to significant errors in the estimates of the viscosity. Direct strain rate measurement is thus beneficial and one possible advantage

FIG. 5. Time history plots at midpoint of superior surface at phonation onset $Q=406 \text{ cm}^3/\text{s}$ (---) and $Q=690 \text{ cm}^3/\text{s}$ (—); (a) inferior-superior displacement, (b) inferior-superior velocity, (c) medial-lateral displacement, and (d) medial-lateral velocity.

of high-speed DIC (as opposed to simpler procedures based on images obtained at low sampling rates). Figure 7(c) shows the medial-lateral strain rate. The strain rate was significantly larger for $Q=690 \text{ cm}^3/\text{s}$ than for Q_{th} . Collision was indicated by a zero strain rate value. The maximum value at $Q=690 \text{ cm}^3/\text{s}$ was nearly twice the mean value. Similar trends were observed for the anterior-posterior strain rates, Fig. 7(d).

Figure 8 shows the distribution of the anterior-posterior ϵ_{yy} and medial-lateral ϵ_{xx} strain components on the deformed superior surface at a flow rate of $Q=690 \text{ cm}^3/\text{s} > Q_C$ in the open and closed states. Figure 8 shows a composite view of the state of deformation along with strain data. In the open state, ϵ_{yy} was tensile for a large central part of the model VF with the maximum at the model VF center, as shown in Fig. 8(a). The region of compressive anterior-posterior strains extended to a distance $L/3$ from the frame edge. During closure, the characteristics of the ϵ_{yy} field changed significantly, Fig. 8(b). At the midsection of the model VFs, the curvature

was reversed and ϵ_{yy} was compressive while tensile strains dominated in the region closer to the enclosure. The maximum tensile value of ϵ_{yy} during closing was well below the maximum tensile strain value observed during opening. In the open state, ϵ_{xx} was compressive for most of the model VF with the minimum at the model VF midsection adjacent to the enclosure, Fig. 8(c). The compressive strain level was reduced near the model VF center. Tensile strains were found only along the surfaces adjacent to the anterior and posterior faces of the enclosure. During closure, the characteristics of the ϵ_{xx} field changed significantly, Fig. 8(d). Most of the superior surface was found to be under tensile medial-lateral strain. These strains were a maximum in the midsection, i.e., the concave domain, with a strain value 30% above that found near VF center, see Fig. 6(b). Asymmetries in the strain fields between the left and right vocal folds are believed to be caused by imperfect placement of the model VFs in the Plexiglas enclosure.

FIG. 6. Strain vs flow rate at midpoint: (a) Anterior-posterior strain ϵ_{yy} and (b) medial-lateral strain ϵ_{xx} . Symbols indicate measured data for open (*) and closed (x) states values. Strain data obtained for $y=0$. $\hat{\epsilon}_{xx}$ is the linear extrapolation of ϵ_{xx} data in the closed state before collision onset.

Contours of stress on the superior surface are shown in Fig. 9. A maximum normalized anterior-posterior stress value of $\sigma_{yy}/G=1.10$ was reached at the model VF midpoint in the open state, Fig. 9(a), while compressive stresses, $\sigma_{yy}/G=-0.25$, were found at the fixed edges. Conversely, for the closed state, Fig. 9(b), $\sigma_{yy}/G=-0.05$ at the VF midpoint and $\sigma_{yy}/G=0.70$ at the fixed edges. Overall, the magnitude of stresses in the anterior-posterior direction was much reduced in the closed state when compared to the open state. In the open state, the normalized medial-lateral stress, Fig. 9(c), possessed its largest tensile value, $\sigma_{xx}/G=0.22$, at the midsection, while compressive stress, $\sigma_{xx}/G=-0.42$, was found near the medial face of the enclosure. In the closed state, Fig. 9(d), values for normalized stress were all positive. Maximum values were $\sigma_{xx}/G=0.44$ at the model VF midsection near the enclosure. The lowest value of the medial-lateral stress, $\sigma_{xx}/G=0.10$, was at the VF center. At this location, the mechanical state of the vocal folds was influenced by collision, as detailed in the following.

C. Hertzian impact model

Of special interest is the mechanical state of the model VFs in the closed state. It has been frequently hypothesized

that the stresses resulting from VF collision are important factors in VF damage, although the exact injury mechanism is unknown. For VF collision to occur, the model VFs need to be stretched along the medial-lateral direction to overcome the glottal gap consisting of the initial prephonatory gap and the gap caused by the deformation due to the mean—or static—pressure on the inferior surface. Evidence of this behavior is found in the contours of medial-lateral strains and stress in the closed state, Figs. 7(d) and 9(d), respectively. The overall tensile deformation state in the medial-lateral direction may seem counterintuitive at first, but it is necessary for collision to occur. Up to the onset of collision, this tensile load state can develop freely without any constraint. Once collision occurs, collision causes new surface stresses and restricts VF motion along the midline.

A Hertzian impact model was used to estimate the collision pressure (Johnson, 2003). Horáček *et al.*, (2005) employed such a model in their computational model based on that of Stronge (2000). Considering rate independence and incompressibility, the normalized collision pressure at the center of the contact area on the medial surface, p_c/G , is given as

FIG. 7. Time history plots at midpoint of superior surface for phonation onset $Q=409 \text{ cm}^3/\text{s}$ (---) and $Q=690 \text{ cm}^3/\text{s}$ (—); (a) anterior–posterior strain, (b) medial–lateral strain, (c) medial–lateral strain rate, and (d) anterior–posterior strain rate.

$$\frac{p_C}{G} = \frac{4\delta_P}{\pi r_{eq}}. \quad (6)$$

In Eq. (6), r_{eq} is the radius of a circular contact region, and δ_P is the (hypothetical) penetration depth of the VFs through the contact plane. It is assumed that impacts occur at speeds that are low relative to the compressive wave speed, and thus are quasistatic (Johnson, 2003). This condition is automatically fulfilled for an incompressible material. The radius of the contact region was obtained from high speed images, as the one in Fig. 3. Contact was presumed to take place over a rectangular area of inferior–superior thickness t_c and anterior–posterior extension $2a$. Since the ellipticity of the contact is only mild [$t_c/(2a) < 2$] the use of a circular contact model with equivalent areas is assumed to introduce only insignificant errors (Greenwood, 2006). The equivalent contact radius r_{eq} was calculated by equating this contact area for the model ($2at_c$) to that of an equivalent circular contact area (πr_{eq}^2), for consistency with Eq. (6) which was originally derived for a circular indenter. The inferior–superior

thickness of the model is $t_c=3.63 \text{ mm}$, see Fig. 1(a). The penetration depth of the model VFs through the contact was estimated from the measured medial–lateral strain at the model VF midpoint. Assuming that in the absence of a second model VF the medial–lateral strain could develop freely, an estimate of the unconstrained deformation of the model VF in that direction was obtained by extrapolation of the initial linear $\epsilon_{xx}^{\text{closed}}-Q$ response, as illustrated in Fig. 6(b). The difference between the extrapolated strain, $\hat{\epsilon}_{xx}$, and the actual measured medial–lateral strain, $\epsilon_{xx}^{\text{closed}}$, provided the contact constraint imposed on VF deformation. For example for $Q=690 \text{ cm}^3/\text{s}$, the difference between the extrapolated strain and the measured strain was $\hat{\epsilon}_{xx}-\epsilon_{xx}^{\text{closed}}=0.053$. The magnitude of the penetration was then obtained using

$$\delta_P = 2(\hat{\epsilon}_{xx} - \epsilon_{xx}^{\text{closed}})l^*, \quad (7)$$

where l^* is the depth of the VF influenced by the contact. The factor two accounts for the presence of the two VFs, assuming symmetry. Combining Eqs. (6) and (7) yielded the

FIG. 8. Contour plots of strains on the deformed superior surface at $Q=690 \text{ cm}^3/\text{s}$: (a), (b) Strain component in the anterior-posterior direction (ϵ_{yy}), and (c), (d) strain component in medial-lateral (ϵ_{xx}) direction, for open and closed states, respectively.

FIG. 9. Contour plots of normalized stress on the deformed superior surface at $Q=690 \text{ cm}^3/\text{s}$: (a), (b) Stress component in the anterior-posterior direction (σ_{yy}), and (c), (d) Stress component in medial-lateral (σ_{xx}) direction, for open and closed states, respectively.

FIG. 10. (a) Equivalent radius, r_{eq} (---), and influence length, l^* (—), vs flow rate; insert shows the estimated influence length for $Q=690 \text{ cm}^3/\text{s}$. (b) Estimated normalized contact pressure, p_c/G , vs flow rate, Q .

collision stress in terms of material properties, measured and extrapolated strains, and the geometry of the contact:

$$\frac{p_c}{G} = \frac{8}{\pi} (\hat{\epsilon}_{xx} - \epsilon_{xx}^{closed}) \frac{l^*}{r_{eq}}. \quad (8)$$

The dependence of the equivalent contact radius r_{eq} on the flow rate is shown in Fig. 10(a) based on measurements of values of $2a$ from high speed images, e.g., Fig. 3(b). Estimates of the influence depth l^* were obtained from the distribution of the medial-lateral stress along the midline ($y = \text{const}$). l^* is the distance from the superior edge to the location on the superior surface where $\partial\sigma_{xx}/\partial x \approx 0$, see insert in Fig. 10(a). The resulting dependence of l^* on flow rate is given in Fig. 10(a). Figure 10(b) then depicts the resulting contact pressure on the medial surface in dependence of the flow rate. The contact pressure was found to vary nonlinearly with flow rate. The contact pressure was zero for flow rates below the collision onset Q_C . For greater flow rates, $Q > Q_C$, the contact pressure increased nonlinearly with flow rate. From the dependence of medial-lateral strains ϵ_{xx}^{closed} on

flow rate [Fig. 6(b)] and of the trends in r_{eq} and l^* [Fig. 10(a)], it can be concluded that the nonlinear response of the contact pressure is due to the saturation of the strain values due to collision.

IV. DISCUSSION AND CONCLUSION

The application of a 3D DIC method for the noncontact and relatively noninvasive (the method requires a paint speckle) determination of the mechanical state of a model vocal fold superior surface during self-oscillation was investigated. The method was successfully implemented and applied in a laboratory experiment using a physical model VF system. The measurements provided time-resolved, full field measurements of several mechanical states of possible interest in phonation studies, including the out-of-plane displacements, the glottal opening displacement, as well as the strain and strain rate fields. Indirect methods were investigated for the estimation of contact pressure from strain data on the superior surface. The use of high speed DIC was beneficial to determine the maximum strain rates directly. Stresses were calculated based on the assumption of incompressible material behavior. The results showed that collision reduced the medial-lateral stress near the region of contact on the superior surface of the model vocal folds. For collision to occur, the model VFs needed to be stretched in the medial-lateral direction in order to overcome the glottal gap. Superimposition of a compressive load due to contact actually reduced stress on the superior surface. Further work is needed to verify these trends, and the accuracy of the contact stress estimates from measured medial-lateral strains and contact radii obtained from high-speed images. The analysis may be biased by the fact that tissue damage, in general, occurs beneath the tissue and not on the superior surface. Therefore, estimates of the magnitude of penetration based on influence depth l^* may be incorrect. The data show that conditions leading to collision also lead to high strain rates, which may be a factor in tissue damage.

The physical and acoustical properties of the physical, synthetic VF model were deemed to adequately represent those found in humans. The vocal fold length L was similar to the upper limit for humans (Zhang *et al.*, 2006). The value of the elastic modulus, assuming incompressibility, was comparable to the lower bound on longitudinal elastic properties of the human VF cover (Zhang *et al.*, 2006; Goodyer *et al.*, 2006). Considering that the longitudinal elastic properties of the vocal fold tissue are considerably higher than that for transverse loading the material employed provides a reasonable homogenized approximation of actual vocal fold tissue stiffness. The frequency of oscillation at onset, $f=88.9 \text{ Hz}$, was at the lower bound of the range of physiological values (Titze, 2000) due to the large vocal fold length. The slow increase of frequency with flow rate beyond onset was consistent with observations for human larynges (Titze, 1989). The characteristics of the physical model during deformation indicated a single orifice was present during the open state, while two smaller, distinct orifices were created around one-quarter span in the closed state. The overemphasis of the third longitudinal mode, along with the absence of a mucosal

wave on the superior and medial surfaces, is not very realistic. These and other similar trends observed in 3D finite element simulations (de Oliveira Rosa *et al.*, 2003) are common to many continuum synthetic vocal folds models. It is important to note that the results obtained with such models are only indicative of general trends, and should not be considered a substitute for data on live tissue. Hopefully, progress in synthetic models along with human subject studies will provide more definitive data in the near future.

The structural differences between the physical model and the human larynx make it difficult to directly relate the results of the present study to actual human phonation. The current model assumed an unstressed initial state, and thus did not account for posturing or pressed phonation. Recognizing these limitations, it is nevertheless interesting to relate the findings of the present study to previous observations made using live tissue models. The estimated influence depth of collision (for flow rates around $Q=690 \text{ cm}^3/\text{s}$) was similar to reported depths of tissue damage (Dikkers and Nikkels, 1995), as well as estimates from computer models (Gunter, 2003). The collision pressures measured in the present study are similar to those reported in hemilarynx experiments. For example Jiang and Titze (1994) found that for a subglottal pressure of around 1 kPa, the peak impact pressure ranged between 0.5 and 1.5 kPa. This compares with an estimated contact pressures around 0.7 kPa for a pressure of 1.2 kPa for the present model. Measurements using pressure sensors inserted between excised canine VFs (Jiang *et al.*, 2001) have yielded wide ranges of contact pressure; for subglottal pressures on the order of 0.9–1.5 kPa, contact pressure values ranged from 0.8–1.5 kPa. These values are slightly higher than those obtained in the present study. FEM models disregarding the flow pressure (Gunter, 2003) have reported compressive stresses orthogonal to the medial surface on the superior surface. Flow studies indicate that fluid pressure acts mainly on the inferior surface. The medial surface is a free surface before collision onset, the stresses orthogonal to that surface are small during the open phase. Upon collision, the contacting medial surfaces are subjected to a compressive load. This contrasts with the tensile stresses measured on the superior surface in the present study. This may be due to the neglect of the fluid pressure loading on the inferior surface in the FEM study or the possibility that both tension and compression exist simultaneously on the superior and medial surfaces during collision, respectively.

This study has demonstrated the potential of digital image correlation analysis for measuring the mechanical state of the superior surface of the vocal folds. Clinical applications are possible, provided methods to safely create a speckle pattern on the superior VF surface are developed. Possible limitations include the need to use only a two-dimensional DIC process, image distortion from endoscopes, glare, and uneven light distributions. The inability to obtain data at discontinuities such as near the glottal gap is also a serious limitation of DIC. Small speckle sizes are needed to partially overcome this problem by allowing for small subset sizes. The potential to assess mechanical states throughout the tissue must be assessed. One possible approach would be the enforcement of displacement data as time varying bound-

ary conditions in finite element models. Optimization techniques may also be used for the indirect determination of material properties in situations where the strain field over the inferior, medial, and superior surfaces could be visible, for example in excised hemi-larynx studies.

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APPENDIX

Reduction of the deviatoric stress vector into a system of linear equations:

$$\underline{\sigma}^d = 2G \left(I_0 - \frac{1}{3} \underline{\underline{mm}}^T \right) \underline{\underline{\epsilon}}, \quad (\text{A1})$$

$$\underline{\sigma}^d = 2G \left(\begin{bmatrix} 1 & 0 & 0 & 0 & 0 & 0 \\ 0 & 1 & 0 & 0 & 0 & 0 \\ 0 & 0 & 1 & 0 & 0 & 0 \\ 0 & 0 & 0 & 1/2 & 0 & 0 \\ 0 & 0 & 0 & 0 & 1/2 & 0 \\ 0 & 0 & 0 & 0 & 0 & 1/2 \end{bmatrix} - \frac{1}{3} \begin{bmatrix} 1 \\ 1 \\ 1 \\ 0 \\ 0 \\ 0 \end{bmatrix} \right) \times \begin{bmatrix} \epsilon_{xx} \\ \epsilon_{yy} \\ \epsilon_{zz} \\ \gamma_{xy} \\ 0 \\ 0 \end{bmatrix}, \quad (\text{A2})$$

$$\underline{\sigma}^d = 2G \left(\begin{bmatrix} 2/3 & -1/3 & -1/3 & 0 & 0 & 0 \\ -1/3 & 2/3 & -1/3 & 0 & 0 & 0 \\ -1/3 & -1/3 & 2/3 & 0 & 0 & 0 \\ 0 & 0 & 0 & 1/2 & 0 & 0 \\ 0 & 0 & 0 & 0 & 1/2 & 0 \\ 0 & 0 & 0 & 0 & 0 & 1/2 \end{bmatrix} \right) \times \begin{bmatrix} \epsilon_{xx} \\ \epsilon_{yy} \\ \epsilon_{zz} \\ \gamma_{xy} \\ 0 \\ 0 \end{bmatrix} = \begin{bmatrix} \sigma_{xx}^d \\ \sigma_{yy}^d \\ \sigma_{zz}^d \\ \tau_{xy} \\ 0 \\ 0 \end{bmatrix}, \quad (\text{A3})$$

$$\frac{\sigma_{xx}^d}{G} = 2[(2/3)\varepsilon_{xx} - (1/3)\varepsilon_{yy} - (1/3)\varepsilon_{zz}], \quad (\text{A4})$$

$$\frac{\sigma_{yy}^d}{G} = 2[-(1/3)\varepsilon_{xx} + (2/3)\varepsilon_{yy} - (1/3)\varepsilon_{zz}], \quad (\text{A5})$$

$$\frac{\sigma_{zz}^d}{G} = 2[-(1/3)\varepsilon_{xx} - (1/3)\varepsilon_{yy} + (2/3)\varepsilon_{zz}]. \quad (\text{A6})$$

Reducing the overall stress vector:

$$\underline{\sigma} = \begin{bmatrix} \sigma_{xx} \\ \sigma_{yy} \\ \sigma_{zz} \\ \tau_{xy} \\ 0 \\ 0 \end{bmatrix} = \underline{\sigma}^d + \sigma^h \underline{m} \approx \underline{\sigma}^d - \sigma_{zz}^d \underline{m} = \begin{bmatrix} \sigma_{xx}^d - \sigma_{zz}^d \\ \sigma_{yy}^d - \sigma_{zz}^d \\ 0 \\ \tau_{xy} \\ 0 \\ 0 \end{bmatrix}. \quad (\text{A7})$$

By incompressibility:

$$\varepsilon_{zz} = -\varepsilon_{xx} - \varepsilon_{yy}. \quad (\text{A8})$$

Inserting Eq. (A8) into Eqs. (A4)–(A6),

$$\frac{\sigma_{xx}^d}{G} = 2[(2/3)\varepsilon_{xx} - (1/3)\varepsilon_{yy} - (1/3)(-\varepsilon_{xx} - \varepsilon_{yy})] = 2\varepsilon_{xx}, \quad (\text{A9})$$

$$\begin{aligned} \frac{\sigma_{yy}^d}{G} &= 2[-(1/3)\varepsilon_{xx} + (2/3)\varepsilon_{yy} - (1/3)(-\varepsilon_{xx} - \varepsilon_{yy})] \\ &= 2\varepsilon_{yy}, \end{aligned} \quad (\text{A10})$$

$$\begin{aligned} \frac{\sigma_{zz}^d}{G} &= 2[-(1/3)\varepsilon_{xx} - (1/3)\varepsilon_{yy} + (2/3)(-\varepsilon_{xx} - \varepsilon_{yy})] \\ &= -2\varepsilon_{xx} - 2\varepsilon_{yy} \end{aligned} \quad (\text{A11})$$

Inserting Eqs. (A9)–(A11) into Eq. (A7) gives the final stress state:

$$\frac{\sigma_{xx}}{G} \approx \frac{(\sigma_{xx}^d - \sigma_{zz}^d)}{G} = 2\varepsilon_{xx} - (-2\varepsilon_{xx} - 2\varepsilon_{yy}) = 4\varepsilon_{xx} + 2\varepsilon_{yy}, \quad (\text{A12})$$

$$\frac{\sigma_{yy}}{G} \approx \frac{(\sigma_{yy}^d - \sigma_{zz}^d)}{G} = 2\varepsilon_{yy} - (-2\varepsilon_{xx} - 2\varepsilon_{yy}) = 4\varepsilon_{yy} + 2\varepsilon_{xx}. \quad (\text{A13})$$

Berry, D. A., Montequin, D. W., and Tayama, N. (2001). "High-speed digital imaging of the medial surface of the vocal folds," *J. Acoust. Soc. Am.* **110**, 2539–2547.

Berry, D. A., Zhang, Z., and Neubauer, J. (2006). "Mechanisms of irregular vibration in a physical model of the vocal folds," *J. Acoust. Soc. Am.* **120**, EL36–EL42.

Chu, C. T., Ranson, W. F., Sutton, M. A., and Peters, W. H. (1985). "Applications of digital-image-correlation techniques to experimental mechanics," *Exp. Mech.* **25**, 232–244.

Dikkers, F. G., and Nikkels, P. G. J. (1995). "Benign lesions of the vocal

folds: Histopathology and phonotrauma," *Manage. Int. Rev.* **104**, 698–703.

de Oliveira, Rosa M., Pereira, J. C., Grellet, M., and Alwan, A. (2003). "A contribution to simulating a three-dimensional larynx model using the finite element method," *J. Acoust. Soc. Am.* **114**, 2893–2905.

Flanagan, J., and Landgraf, L. (1968). "Self-oscillating source for vocal-tract synthesizers," *IEEE Trans. Audio Electroacoust.* **16**, 57–64.

Fulcher, L. P., Scherer, R. C., Zhai, G., and Zhu, Z. (2006). "Analytic representation of volume flow as a function of geometry and pressure in a static physical model of the glottis," *J. Voice* **20**, 489–512.

Gardner, G. M., Castracane, J., Conerty, M., and Parnes, S. M. (1995). "Electronic speckle pattern interferometry of the vibrating larynx," *Ann. Otol. Rhinol. Laryngol.* **104**, 5–12.

Goodyer, E., Muller, F., Bramer, B., Chauhan, D., and Hess, M. (2006). "In vivo measurement of the elastic properties of the human vocal fold," *Eur. Arch. Otorhinolaryngol.* **263**, 455–462.

Greenwood, J. A. (2006). "A simplified elliptic model of rough surface contact," *Wear* **261**, 191–200.

Gunter, H. E. (2003). "Mechanical stress in vocal fold tissue during voice production," Ph.D. thesis, Division of Engineering and Applied Sciences, Harvard University, Cambridge, MA.

Helm, J. D., McNeill, S. R., and Sutton, M. A. (1996). "Improved three-dimensional image correlation for surface displacement measurement," *Opt. Eng. (Bellingham)* **35**, 1911–1920.

Horáček, J., Šidlof, P., and Švec, J. G. (2005). "Numerical simulation of self-oscillations of human vocal folds with Hertz model of impact forces," *J. Fluids Struct.* **20**, 853–869.

Hung, P., and Voloshin, A. S. (2003). "In-plane strain measurement by digital image correlation," *J. Braz. Soc. Mech. Sci.* **25**, 215–221.

Ishizaka, K., and Flanagan, J. L. (1972). "Synthesis of voiced sounds from a two-mass model of the vocal folds," *Bell Syst. Tech. J.* **51**, 1233–1268.

Jiang, J. J., Shah, A. G., Hess, M. M., Verdolini, K., Banzali, F. M., Jr., and Hanson, D. G. (2001). "Vocal fold impact stress analysis," *J. Voice* **15**, 4–14.

Jiang, J. J., and Titze, I. R. (1994). "Measurement of vocal fold intraglottal pressure and impact stress," *J. Voice* **8**, 132–144.

Johnson, K. L. (2003). *Contact Mechanics* (Cambridge University Press. Cambridge, UK), Chap. 11.

Kerdok, A. E., Cotin, S. M., Ottensmeyer, M. P., Galea, A. M., Howe, R. D., and Dawson, S. L. (2003). "Truth cube: Establishing physical standards for real time soft tissue simulation," *Med. Image Anal.* **7**, 283–291.

Manneberg, G., Hertegard, S., and Liljencrantz, J. (2001). "Measurement of human vocal fold vibrations with laser triangulation," *Opt. Eng. (Bellingham)* **40**, 2041–2044.

Mantha, S., Mongeau, L., and Siegmund, T. (2005a). "Dynamic digital image correlation of a dynamic physical model of the vocal folds," *Adv. Bioeng.* **57**, 77–78.

Mantha, S., Siegmund, T., and Mongeau, L. (2005). "Estimation of strain fields in self oscillating physical glottis models using 3D digital image correlation," *Proceedings of the Fourth International Workshop on Models and Analysis of Vocal Emissions for Biomedical Applications*, Florence, Italy, 29–31 October.

Mongeau, L., Franchek, N., Coker, C., and Kubli, R. (1997). "Characteristics of a pulsating jet through a small modulated orifice, with application to voice production," *J. Acoust. Soc. Am.* **102**, 1121–1133.

Ouaknine, M., Garrel, R., and Giovanni, A. (2003). "Separate detection of vocal fold vibration by optorelectometry: A study of biphonation on excised porcine larynges," *Folia Phoniatr Logop* **55**, 28–38.

Ouaknine, M., Garrel, R., and Giovanni, A. (2003). "Separate detection of vocal fold vibration by optorelectometry: A study of biphonation on excised porcine larynges," *Folia Phoniatr Logop* **55**, 28–38.

Park, J. B., and Mongeau, L. (2007). "Instantaneous orifice discharge coefficient of a physical, driven model of the human larynx," *J. Acoust. Soc. Am.* **121**, 442–455.

Scherer, R. C., Shinwari, D., De Witt, K. J., Zhang, C., Kucinski, B. R., and Afjeh, A. A. (2001). "Intraglottal pressure profiles for a symmetric and oblique glottis with a divergence angle of 10 degrees," *J. Acoust. Soc. Am.* **109**, 1616–1630.

Schreier, H. W., Garcia, D., and Sutton, M. A. (2004). "Advances in light microscope stereo vision," *Exp. Mech.* **44**, 278–288.

Siebert, T., Becker, T., Spillthof, K., Neumann, I., and Krupka, R. (2007). "High-speed digital image correlation: Error estimations and applications," *Opt. Eng.* **46**, 051004-1–7.

Spencer, M. (2007). "Indirect determination of the strain and stress in physi-

- cal models of the vocal folds using digital image correlation," Masters thesis, Purdue University West Lafayette, IN.
- Stronge, W. L. (2000). *Impact Mechanics* (Cambridge University Press, Cambridge, UK), Chap. 6, pp. 116–127.
- Sutton, M. A., Cheng, M., Peters, W. H., Chao, Y. J., and McNeil, S. R. (1986). "Application of an optimized digital image correlation method to planar deformation analysis," *Image Vis. Comput.* **4**, 143–150.
- Sutton, M. A., Helm, J. D., and Boone, M. L. (2001). "Experimental study of crack growth in thin sheet 2024-T3 aluminum under tension-torsion loading," *Int. J. Fract.* **109**, 285–301.
- Švec, J. G., Horáček, J., Sram, F., and Vesely, J. (2000). "Resonance properties of the vocal folds: in vivo laryngoscopic investigation of the externally excited laryngeal vibrations," *J. Acoust. Soc. Am.* **108**, 1397–1407.
- Tao, C., Jiang, J. J., and Zhang, Y. (2006). "Simulation of vocal fold impact pressure with a self-oscillating finite-element model," *J. Acoust. Soc. Am.* **119**, 3987–3994.
- Thomson, S. L., Mongeau, L., and Frankel, S. H. (2005). "Aerodynamic transfer of energy to the vocal folds," *J. Acoust. Soc. Am.* **118**, 1689–1700.
- Tigges, M., Wittenberg, T., Rosanowski, F., and Eysholdt, U. (1996). "High-speed imaging and image processing in voice disorders," *Proc. SPIE* **2927**, 209–216.
- Titze, I. R. (1989). "On the relation between subglottal pressure and fundamental frequency in phonation," *J. Acoust. Soc. Am.* **85**, 901–906.
- Titze, I. R. (1994). "Mechanical stress in phonation," *J. Voice* **8**, 99–105.
- Titze, I. R. (2000). *Principles of Voice Production* (National Center for Voice and Speech, Iowa City, IA).
- Verdolini, K., Hess, M. M., Titze, I. R., Bierhals, W., and Gross, M. (1999). "Investigation of vocal fold impact stress in human subjects," *J. Voice* **13**, 184–202.
- Zhang, K., Siegmund, T., and Chan, R. W. (2006). "A constitutive model of the human vocal fold cover for fundamental frequency regulation," *J. Acoust. Soc. Am.* **119**, 1050–1062.
- Zienkiewicz, O. C., and Taylor, R. L. (2000). *The Finite Element Method* (Butterworth-Heinemann, Oxford, UK), Chap. 12.