A computational study of depth of vibration into vocal fold tissues

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The effective depth of vocal fold vibration is self-regulated and generally not known *a priori* in vocalization. In this study, the effective depth was quantified systematically under various phonatory conditions using a fiber-gel finite element vocal fold model. The horizontal and vertical excursions of each finite element nodal point trajectory were recorded to compute trajectory areas. The extent of vibration was then studied based on the variation of trajectory radii as a function of depth in several coronal sections along the anterior-posterior direction. The results suggested that the vocal fold nodal trajectory excursions decrease systematically as a function of depth but are affected by the layered structure of the vocal folds. The effective depth of vibration was found to range between 15 and 55% of the total anatomical depth across all phonatory conditions. The nodal trajectories from the current study were compared qualitatively with the results from excised human hemi-larynx experiments published in Döllinger and Berry [(2006). *J. Voice*, **20**(3), 401–413]. An estimate of the effective mass of a one-mass vocal fold model was also computed based on the effective depth of vibration observed in this study under various phonatory conditions.

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I. INTRODUCTION

Tissue motion in vocal fold vibration is routinely investigated with video endoscopy (Mehta *et al.*, 2011; Deliyski *et al.*, 2008). While much detail is available on surface motion, only a few studies have addressed the motion of tissue particles deep inside the vocal folds (Saito *et al.*, 1981; Fukuda *et al.*, 1983). If the tissue properties were homogeneous, the internal movement would be predictable from surface movement on the basis of stress and strain continuity within the tissue. However, the layered structure of the vocal folds can cause sudden discontinuities in movement. The ligament may move differently from the superficial layer or the muscle. Furthermore, only a portion of the muscle may be in vibration. Thus, the effective depth of vibration (medial to lateral) into the vocal fold is always in doubt. For vocal fold modeling with low-dimensional (mass-spring) models, the mass is estimated from the length, thickness, and depth of the effective tissue in motion (Titze and Story, 2002). Given that the depth is always an estimate, not based on measurement or accurate prediction, the mass in vibration is generally unknown. When surgical medialization of a paralyzed vocal fold is contemplated with injectable or implantable material that is too stiff to vibrate (Ryu *et al.*, 2012), there is always a question of how close to the medial surface the

material can be placed without disturbing the vibration. In this study, finding the effective depth of vocal fold vibration under various phonatory conditions was explored through a series of experiments using a finite element fiber-gel vocal fold model (Titze *et al.*, 2017).

Previously, Saito *et al.* (1981) conducted an X-ray stroboscopy study with implanted lead pellets in excised human and canine larynges. In one part of the study, one pellet was in the superficial layer of the lamina propria, one pellet was in the deep layer of the lamina propria, and one was in the muscle. The muscle pellet had the smallest amplitude of vibration, indicating that vibration diminishes with depth into the tissue. While pellet movement was successfully tracked and the trajectories were well-defined circles or ellipses, the pellets added a mechanical load to the tissue due to their much higher density (a factor of 11 greater than tissue). This mechanical load brought into doubt the accuracy of the overall vocal fold movement. Nevertheless, it was shown that the radii of circular or elliptical trajectory movement were smaller with greater depth into the tissue.

In a subsequent study by the same group (Fukuda *et al.*, 1983), lead pellets were also implanted in a live canine vocal fold in which thyroarytenoid muscle activation could be varied with recurrent nerve stimulation. The vibrational amplitude in the muscle was reduced with nerve stimulation, suggesting that muscle stiffness prevents large-amplitude vibration deep into the tissue. More data were collected with this paradigm by Saito *et al.* (1985). It was again shown that small amplitude of vibration occurs deep in the muscle layer. The reduction in the area of the trajectory was more than ten-fold in comparison to the trajectory area in the superficial layer.

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Alipour-Haghighi and Titze (1985) simulated particle trajectories with the first version of a finite element model of the vocal folds. The mesh in the coronal plane was coarse and there was no layered tissue structure. For this homogeneous tissue construct, the vertical movement was significantly greater than lateral movement in a trajectory. A gradual linear progression in amplitude reduction took place medial to lateral, but the deepest element node tracked was less than half of the tissue depth to the lateral (thyroid) boundary. Thus, the nature of the amplitude decay was poorly defined. Deguchi *et al.* (2011) also used finite elements to study the effective depth of vocal fold vibration. However, they analyzed the effective depth only indirectly by investigating the effect of thyroarytenoid (TA) and cricothyroid (CT) muscle activations on vocal fold morphology and stress distribution.

On the surface of the vocal folds, particle trajectory movement was first quantified by Baer (1975) on excised larynges. Fine metal particles were sprinkled onto the surface of the vocal folds and tracked with X-ray imaging. An excised hemi larynx approach was later used by Berry *et al.* (2001) on canine vocal folds. Nine fleshpoints were tracked simultaneously along the medial surface of one coronal plane of the left vocal fold. The distance between microsutures was approximately 1 mm. The study reported an anterior view of the elliptical paths in two-dimensional (2D) plane traversed by the fleshpoints over one cycle of vibration. The vibration pattern had a fundamental frequency of 102 Hz and the trajectories were predominantly circular or elliptical. The radii of the fleshpoint trajectories decreased significantly from superior to inferior location.

Using human larynges, Döllinger and Berry (2006) performed the excised hemi larynx study on one larynx. The dynamics of 35 microsutures mounted on the medial surface of a human vocal fold were analyzed across 18 phonatory conditions. The microsutures each had a diameter of 0.034 mm. They were placed along the medial surface in a 7×5 grid. The vertical distance between the sutures was approximately 1.7 mm and the horizontal distance was approximately 2 mm. The suture movements within the recorded time period were reported. The measured fundamental frequencies were in the speech range between 115 and 140 Hz. Döllinger *et al.* (2016) repeated the *ex vivo* hemilarynx experiment in three human larynges to visualize the displacement of medial vocal fold surface. The dynamics of 30 microsutures sewed into the mucosal epithelium were tracked across 30 experiments. The distance between two neighboring sutures was 1.96 ± 0.3 mm. Fundamental frequency was found to be between 97 and 200 Hz.

The purpose of the current study was to quantify the effective depth of vibration with an improved finite-element computational model of the vocal folds. The model allows effective depth to be obtained without adding lead pellet loads or microsutures on the vocal folds. The fleshpoint and microsuture displacement results from the Döllinger and Berry (2006) *ex vivo* human hemi-larynx study were used for qualitative comparison with our computations. The primary questions for this study were: (1) do the trajectory amplitudes of tissue particles vary linearly from the medial surface to the lateral boundary? (2) How does the depth of vibration vary with lung pressure? (3) Is depth of vibration affected by

thyroarytenoid and cricothyroid muscle contraction? And (4) is depth of vibration uniform along the thickness and length of the vocal folds?

II. METHODS

A. Fiber-gel finite element model

The fiber-gel vocal fold model of the computational software *VoxInSilico* (Titze *et al.*, 2017; Titze *et al.*, 2014) was used as the investigative tool. The basic geometrical shape of the vocal folds was based on the general anatomical data for humans (Titze, 2006, Chapter 2). For the current study, the length of the vocal folds was 1.6 cm, an average for adult human males. The vertical thickness on the medial and lateral surfaces was 0.7 and 1.5 cm, respectively, which included the subglottal taper in the conus elasticus region (Fig. 1). The anatomical depth of the vocal folds, defined in the medial-lateral dimension, varied in the anterior-posterior direction, and followed the curvature of the thyroid cartilage boundary. For pre-phonatory posturing, the anatomical dimensions of the vocal folds were modified by activation of the TA and CT muscles. This modification was accomplished by empirical equations (rules) that approximate the framework mechanics (Titze, 2006, Chapter 3). In the transverse plane, the horizontal (x , y) positions of the tips of the vocal process of the arytenoid cartilage were determined by vector forces of translation and rotation of the thyroid cartilage around the cricothyroid joint. For vocal fold length, an empirical rule that approximated vocal fold strain (fractional elongation) was

$$\varepsilon = G(Ra_{CT} - a_{TA}) - Ha_{LCA}, \quad (1)$$

where $G = 0.3$, $R = 2$, and $H = 0.4$ are empirical constants obtained based on experiments with excised larynges and *in vivo* animal studies (Titze *et al.*, 1988; Titze *et al.*, 1997). The parameters a_{TA} , a_{CT} , and a_{LCA} are activation levels of TA, CT, and lateral cricoarytenoid (LCA) muscles,

FIG. 1. (Color online) Coronal section of the right vocal fold. Purple is the mucosa layer, yellow is the ligament, and red is the TA muscle. Notation for the nodes is Node (row, column).

respectively. For example, when TA activation was at 5% and CT activation at 30%, the steady state vocal fold strain was 4%. At these activation levels, the anatomic depth of the vocal folds varied from 0.21 cm at the anterior end to 1.57 cm at the posterior end. On the basis of tissue incompressibility, the thickness of the vocal folds varied inversely with strain

$$T = T_0/(1 + \varepsilon), \quad (2)$$

where T_0 is the resting thickness. The full detail of vocal fold posturing and shape modifications with intrinsic laryngeal muscle activation is given in [Titze \(2006, Chapter 3\)](#).

Each vocal fold had three layers: a superficial layer of the lamina propria (SLLP) including the epithelium, a ligament, and a muscle. The viscoelastic properties of these layers were also (in part) governed by muscle activations. Each layer has a shear modulus μ in a plane perpendicular to the fibers and a shear modulus μ' in the plane of the fibers. The transverse (gel) shear moduli $\mu = 0.5$ kPa for all the three layers are currently not dependent on muscle activation, but the longitudinal (fiber) shear moduli μ' vary with ε , which is based on muscle activation [Eq. (1)]. The tissues are considered transversally isotropic in the coronal plane. The equation governing longitudinal shear moduli μ' is given as

$$\mu' = \mu'_o + T/dxdz = \mu'_o + \sigma_y. \quad (3)$$

Here, μ'_o is the shear modulus in a plane perpendicular to the transverse plane and is currently set equal to μ . T is the fiber tension, and $dxdz$ is the fiber cross section in the transverse plane. The details of this fiber-gel characterization are given in [Titze et al. \(2017\)](#). The fiber stress σ_y varies with vocal fold strain, ε and the governing equation along with their constants is given in [Titze \[2006, Chapter 2, Eq. \(2.41\)\]](#).

A 7×6 mesh was used in the coronal plane, which allowed the trajectories of eight nodal points (including the boundaries) to be tracked in the medial-lateral direction and seven nodal points in the superior-inferior direction (Fig. 1). Fifteen coronal slices were used in the anterior-posterior direction, resulting in a $7 \times 15 \times 6$ mesh. The tissue was divided into triangular elements in the coronal plane and rectangular elements along the length in the anterior-posterior direction. The mesh resolution allowed tracking the trajectories on the medial surface with a nodal point distance of 1.07 mm along the anterior-posterior length and 1.16 mm along the superior-inferior thickness.

B. Data acquisition

The horizontal (x) and vertical (z) excursions of each nodal point trajectory in the coronal plane were calculated in terms of trajectory area. This calculation was conducted in three of the 15 coronal slices in the anterior-posterior (y) direction. From the slices indexed 1:15, the ones selected for particle trajectories were 2, 8, and 14. This provided data at 10, 50, and 90% along the length of the membranous vocal fold. This variation with anterior-posterior (A-P) position needed to be included because the anatomical depth of the tissue varied due to the curvature of the thyroid cartilage boundary. The depth of vocal fold vibration was computed

for several lung pressures (P_L), TA activation, and CT activation levels. Lung pressure was varied between 0.5 and 2.5 kPa, TA activation was varied between 5 and 100%, and CT activation was varied between 30 and 100%. LCA, interarytenoid (IA), and posterior cricoarytenoid (PCA) activations were kept constant at 30, 30, and 0%, respectively, across all simulations. These nominal muscle activation values were chosen such that they provide fundamental frequencies in the speech range ([Titze et al., 1989](#)). In addition, the choice of keeping LCA activation at 30% is consistent with the work of [Chhetri and Neubauer \(2015\)](#), which showed that LCA activation levels in the mid-range is synergistic with TA activation. Since IA acts in concert with LCA in adduction, and their innervating nerve branches are in close proximity and typically stimulated together in *ex vivo* experiments ([Chhetri et al., 2014](#)), it also makes sense to fix IA activation level to be the same as LCA.

C. Data analysis

The areas A of steady-state vocal fold trajectories were calculated by creating a single conformational 2D boundary around the (x, z) excursions of each nodal point and measuring the area which the boundary encloses. The effective trajectory radius was computed as $\sqrt{A/\pi}$. The *effective depth of vocal fold vibration* was defined as the depth at which the area under the trajectory radius curve as a function of anatomic depth was half of the total area (medial to lateral). This definition of effective depth follows the logic that, if vibrational amplitude $a(x)$ is maximum at the medial surface and zero at the fixed lateral boundary, the mean amplitude would be proportional to the integral of $a(x)dx$, an average area value. Such a definition is equivalent to effective skin depth or skin effect in electromagnetics, which is measured as an approximation for field penetration in a conductor ([Wentworth et al., 2006](#)). The skin depth was defined with the hypothesis that the current density is highest at the surface and decreases exponentially with increasing depth into a homogeneous conductor ([Wheeler, 1942](#)). Hence, the skin depth is quantified as the depth into the conductor at which the current density reaches 37% ($1/e$) of the value at the surface. However, vocal fold tissue is not homogeneous and the vibration decay is not exponential or linear (as shown later). As a result, we do not use the definition of 37% of the value at the surface to represent effective depth. Instead, in this study, we use the definition of mean area under the trajectory radius curve, i.e., the location that divides the area under the curve into two equal halves. Figure 2 shows an example trajectory radius curve with its effective depth labeled in terms of the percentage of the total anatomical depth. The area under the curve to the left and right of the effective depth location is equal.

III. RESULTS

Figure 3 shows a matrix of coronal plane views of the right vocal fold with variable TA activation (columns) and variable lung pressure (rows). The coronal plane shown is the center slice (index 8) along the 15 slices in the anterior-posterior direction. Trajectories are shown for an 8×7 grid

FIG. 2. (Color online) An example trajectory radius curve with a location of effective depth.

of nodal points, including nodes at the boundaries. The trajectories are shown for the entire duration of the simulation, which includes both transient and steady-state vocal fold oscillation. The range of fundamental frequencies covered

with these 25 simulations was 109–159 Hz, all basically in the male speech range with an average of 121 Hz. Note that most of the trajectories were ellipses, with x and z displacements being similar. The trajectories were more circular at the lower medial surface. The upper and medial-most nodes had the greatest excursions and were the flattest ellipses. The areas of the trajectories decreased from superior to inferior as well as from medial to lateral. This finding of a decrease in the trajectory areas from superior to inferior direction was also observed in [Berry *et al.* \(2001\)](#) study on canine vocal folds. Steady, self-sustained vibration was observed for TA activation $\leq 50\%$ except in the upper left case. In the upper left (TA is 5%, and P_L is 0.5 kPa) case, there was no vibration because the combination of 0.5 kPa pressure and 5% TA activation was below the threshold for phonation. For large values of TA activation (75% and above), some very complex trajectories were seen at the medial surface, but the majority of them consisted only of a transient response. There was no steady-state oscillation at very high TA activation levels due to stiff muscle resulting in damped oscillation.

Figure 4 shows the steady-state nodal point trajectory areas of different nodes as a function of lung pressure for different TA activation levels. The areas shown were from the coronal section at the center (index 8) along the anterior-posterior direction. There was a uniform increase in the areas of the trajectories with increasing lung pressure. The relationship is almost linear for all the nodes. The maximum nodal point trajectory areas were obtained for TA activation of 25%, followed by 50%, and 5%. Note that there is a difference in the scales along the y-axis for each node. A significant decrease [$p < 0.001$ based on one-way analysis of variance (ANOVA)] can be seen in the areas from one column to the next. There was also a decrease in the areas from one row to the next ($p < 0.01$ based on one-way ANOVA).

Figure 5 shows the steady-state nodal point trajectories' areas as a function of TA activation levels for different lung pressures, the same data as in Fig. 4, but a reversal of the x axis variable and the curve parameter. The maximum excursions (and trajectory areas) occurred with TA activation on

FIG. 3. Nodal point trajectories for a matrix of coronal sections of the right vocal fold. TA activation increases from 5% (left) to 100% (right) and lung pressure increases from 0.5 kPa (top) to 2.5 kPa (bottom). CT activation remained at 30%. Fundamental frequencies were in the speech range (109–159 Hz).

FIG. 4. (Color online) Steady-state nodal point trajectories' areas in the mid-coronal section along A-P direction as a function of lung pressure for different TA activation levels. The notation for nodes within a coronal slice is Node (row, column). Note the difference in the scales along the y-axis for each node.

the order of 5%–50% for all nodes with the highest at 25% TA activation. A linear relationship with lung pressure can be observed even in Fig. 5. There was no steady-state vocal fold oscillation at 100% TA activation for any lung pressure and at 75% TA activation for lung pressures below 1.5 kPa.

Figure 6, left to right, shows examples of how the trajectories change in the posterior-anterior (P-A) direction. Slices were taken along the vocal fold length at 10% from the posterior boundary, center, and 10% from the anterior boundary.

The examples were taken from the bottom of the P_L -TA matrix in Fig. 3 (row 5, column 2), where $P_L = 2.5$ kPa and TA = 25%, a case with the highest amplitude of vibration. The CT activation for the top row was the nominal 30%. Note that the trajectories at both posterior and anterior ends are substantially smaller than those at the mid-membranous slice.

The lower part of the figure shows the results for an increase in CT activation, from 30% to 80%. This increased

FIG. 5. (Color online) Steady-state nodal point trajectories' areas for the mid-coronal section along A-P direction as a function of TA activation levels for different lung pressures. The notation for nodes within a coronal slice is Node (row, column). Note the difference in the scales along the y-axis for each node.

FIG. 6. Variation in the nodal point trajectories for a matrix of coronal plane sections of the right vocal fold. From left to right, the trajectories are for sections 10% from the posterior boundary, center, and 10% from the anterior boundary. From top to bottom, the CT activation is 30 and 80%, respectively.

the fundamental frequency to 340 Hz. The trajectories diminished substantially in all three sections, especially in the superior-medial corner where the ligament resides. Note the restricted motion of nodes that cover the ligament compared to the mucosa and muscle layer nodes in the center coronal section. At 80% CT activity, the ligament is very stiff.

The steady-state trajectory areas for the nominal (median value) case were listed in Table I. They range from 0.0012 mm^2 in node (7,7) to 1.0484 mm^2 in node (1,1). It can be observed that the trajectory areas decrease consistently along the medial-lateral as well as superior-inferior direction with a few exceptions. Figure 7 shows a plot for the equivalent radius $\sqrt{A/\pi}$ as a function of depth (similar to Fig. 2) at different rows along the superior-inferior direction for the test cases in Fig. 6. At 30% CT activation, the radius decreased not quite linearly, but also not exponentially. At nodal point 4, just behind the ligament, the equivalent radius is reduced to about half of the value at the medial surface. The variation in the superior-inferior (z) direction is

TABLE I. Areas in mm^2 for 8×7 nodal point trajectories for median value parameters, $P_L = 1.5 \text{ kPa}$, $TA = 50\%$, $CT = 30\%$, and the mid-membranous A-P position. Medial to lateral nodes are notated along the top (1–8), and superior to inferior nodes are notated down the leftmost column (1–7).

	1	2	3	4	5	6	7	8
1	1.0484	0.5094	0.3025	0.2304	0.1497	0.0590	0.0172	0.0000
2	0.8692	0.5177	0.3521	0.2069	0.0814	0.0230	0.0043	0.0000
3	0.7547	0.4590	0.2433	0.0923	0.0328	0.0121	0.0040	0.0000
4	0.5071	0.2434	0.0896	0.0410	0.0214	0.0092	0.0029	0.0000
5	0.1596	0.0573	0.0333	0.0247	0.0140	0.0063	0.0022	0.0000
6	0.1984	0.0625	0.0208	0.0146	0.0083	0.0040	0.0016	0.0000
7	0.2737	0.0758	0.0137	0.0075	0.0042	0.0022	0.0012	0.0000

significant at the medial surface but there is not much difference in the equivalent radius towards the lateral boundary. The trajectory radius decreased drastically at 80% CT activation level. The effective radius is also lower at the second to fourth nodes where the ligament is present compared to the mucosa and muscle regions. The trajectory radius is higher at the superior margin compared to the inferior margin, as well as the medial margin compared to the lateral margin, as observed in Fig. 3 and Fig. 6.

Figure 8 shows the effective depth of vibration for the test cases in Fig. 3 at different rows along the superior-to-inferior direction. Only the cases where self-sustained oscillation occurred were selected for the analysis of the effective depth of vocal fold vibration (i.e., $P_L \geq 1.0 \text{ kPa}$, and $TA \leq 50\%$). The effective depth of vibration was found to be between 15% and 45% of the anatomical depth across all the cases. It can be observed that the effective depth varied only slightly with lung pressure, but decreased with an increase in TA activation level. The effective depth is higher at the superior end compared to the inferior end for all the cases and decreased linearly for the majority of cases. There were a few cases at 5% TA activation where the effective depth is higher at the center compared to the edges along the superior-inferior direction.

Figure 9 shows the effective depth for the cases shown in Fig. 6. The effective depth was found to be between 20% and 55% of the anatomical depth of the vocal fold. From the superior to the inferior end, the effective depth of vibration decreased for 30% CT activation and increased for 80% CT activation. From the posterior to the anterior end, the effective depth increased for 30% CT level and decreased for 80% CT level. The effective depth of vibration is greater at higher CT level (close to 50% or above) compared to the lower CT level, but the trajectory radius is very small.

A. Comparison with measurement

The trajectories obtained in this study were compared qualitatively with the X-ray and optical data from the Döllinger and Berry (2006) study. Experimental conditions were different between the two studies to perform any meaningful quantitative analysis. For comparison purposes, we chose the lung pressure to be 1.5 kPa, TA activation of 25%, and CT activation of 30%. The remaining parameters were at their nominal values. In the Döllinger and Berry (2006) study, the subglottal pressure was approximately 2.8 kPa. They attached weights to the arytenoid and thyroid cartilages to vary the vocal fold posture instead of intrinsic laryngeal muscle activations. Their study also does not include supraglottal vocal tract. Microsuture trajectories on the medial surface from an *ex vivo* human larynx were presented in Fig. 6 of the Döllinger and Berry (2006) paper. They presented the trajectories at six different vertical (superior-to-inferior) sections along the anterior (C1)-posterior (C5) direction. We computed the trajectory radius of those micro-suture excursions by extracting data from their published figure. Figure 10(A) shows the trajectory radius as a function of superior-inferior thickness for the five sections along the A-P direction from the Döllinger and Berry (2006) study. Similarly, Fig. 10(B)

FIG. 7. (Color online) Equivalent radius of trajectories as a function of depth for CT activation of 30% (top row) and 80% (bottom row). Row 1 is at the superior end and row 7 is at the inferior end. First column of sub-figures corresponds to coronal sections at the posterior end, the second column to coronal sections at the center, and third column to coronal sections at the anterior end. Note that the scale along the y-axis is different between the top and bottom rows.

FIG. 8. Effective depth of vibration in percentage of the anatomical depth along the superior-to-inferior direction for different TA activation and P_L levels.

FIG. 9. The effective depth of vibration as a percentage of the anatomical depth along the superior-to-inferior direction for different CT activation levels.

shows the trajectory radius as a function of thickness from the fiber-gel finite element vocal fold model. The C1-C5 coronal sections corresponds to 10, 30, 50, 70, and 90% from the anterior end, respectively.

In both studies, it can be observed that the trajectory radius decreased for all five sections from superior to inferior surfaces. At the superior surface, the trajectory radius is higher at the center along the A-P direction compared to the edges. In the fiber-gel model, the minimum radius occurred at the center along the superior-inferior direction, suggesting a (1,1) mode of vibration. This was unlike the minimum at the inferior surface observed in the Döllinger and Berry (2006) study. The lack of

(1,1) mode of vibration in the Döllinger and Berry (2006) study could be due to different driving pressures. Also, a difference in the trajectory radius along A-P direction was observed at the superior surface but not at the inferior surface in the Döllinger and Berry (2006) study. In contrast, a significant difference in the trajectory radius was observed along the A-P direction at both the superior and inferior surfaces with the fiber-gel model. The trajectory radius amplitude is higher at the center along the A-P direction in the fiber-gel model compared to the results from Döllinger and Berry (2006) study, despite the model having a significantly lesser lung pressure. This could be due to the difference in experimental conditions between the two studies.

FIG. 10. (Color online) (A) Trajectory radius in cm for nodal points along thickness on the medial surface calculated from data presented in the Döllinger and Berry (2006) paper. (B) Trajectory radius in cm for nodal points along thickness on the medial surface obtained from Fiber-Gel model. C1 is a column at the anterior end and C5 is a column at the posterior end.

IV. DISCUSSION AND CONCLUSIONS

In this study, the effective depth of vocal fold vibration was systematically studied under various phonatory conditions using a fiber-gel finite element vocal fold model. It was observed that the trajectory radius decreased from superior-to-inferior direction and also from medial-to-lateral direction at lower CT activation levels. At higher CT activation levels, the trajectory radius decreased until the level of ligament and increased slightly into the TA muscle region along the medial-lateral direction. The amplitude of vocal fold vibration decreased drastically with increase in CT activation level due to stiff ligament. The vocal fold vibration increased almost linearly with an increase in lung pressure. It was highest at 25% TA activation and did not oscillate at TA activation levels above 75% due to the stiffness of the muscle.

The effective depth was defined based on relative amplitude of vibration along depth rather than absolute amplitude at the surface. This is because the tissue is multi-layered instead of spatially uniform. The effective depth depends on how well energy is coupled to the fibers that dominate in vibration and which layer determines the frequency and mode of vibration. In such multi-modal systems, there are always local peaks and valleys of amplitude of vibration. Thus, absolute amplitude at any specific location is not a predictor of overall effective depth of vibration.

The effective depth of vocal fold vibration was found to be between 15 and 55% of the anatomical vocal fold depth for various combinations of lung pressure, CT, and TA muscle activations. Lung pressure did not have much of an effect on effective depth of vocal fold vibration. The effective depth appeared to decrease with an increase in TA activation level (Fig. 8) and increase with an increase in CT level (Fig. 9). At low CT level, the effective depth of vocal fold vibration was found to be higher at the superior margin compared to the inferior margin for all the test cases. The opposite was true at higher CT level. The muscle is stiff in relation to the ligament at the lower CT level, whereas ligament is stiff compared to muscle at higher CT level. Hence, at CT activation of 30%, the effective depth decreased in the superior-to-inferior direction, whereas at 80% CT level, it increased in the superior-to-inferior direction. The CT activation level also had similar effect on effective depth along the anterior-posterior direction (Fig. 9). The effective depth decreased at lower CT level and increased at higher CT level along the anterior-posterior direction.

The simulation shows that, for low-frequency (speech-like) vocal fold oscillation, the effective depth of vibration (medial to lateral) is on the order of half the anatomical depth measured from the medial surface to the thyroid cartilage. This suggests that much of the thyroarytenoid muscle is effectively only partially in vibration, especially the thyromuscularis (lateral) portion. This may be the reason that muscle fiber orientation is not highly specific in this muscularis region (Titze and Strong, 1975), unlike in the thyrovocalis (medial) region where fibers run mostly in the anterior-posterior direction (Cox *et al.*, 1999). The implication is that structural and viscoelastic properties are free to be modified in this lateral portion of the vocal folds, as phonosurgeons

have discovered with type 1 thyroplasty (Isshiki *et al.*, 1974) and injection augmentation (Ford and Bless, 1986). For fundamental frequencies higher than 350 Hz (not tested in this study), the effective depth is expected to be considerably smaller, not reaching beyond the vocal ligament.

Results of the simulation were compared qualitatively with experimental data. Vocal fold vibration was observed to be greater at the center compared to the posterior or anterior ends, consistent with Döllinger and Berry (2006) study. Along the medial surface, the trajectory radius decreased from the superior to inferior margin in the Döllinger and Berry (2006) data. In the simulation, the trajectory radius was larger at the superior margin compared to the inferior margin for all the sections along the A-P direction, consistent with experimental data. However, the trajectory radius was smallest halfway along the superior-inferior direction compared to the edges. This suggests a (1,1) mode of vibration with a nodal point near the center in the fiber-gel model.

The results of this work have implications for further refinement of the fiber-gel finite element computational model. The finite-element size (spatial discretization) can be related inversely to the particle trajectory area to capture the detail of vibration in the most economical way. In other words, smaller elements can be used medially and larger elements laterally. This feature was adopted in the finite-element model of Alipour *et al.* (2000), but it was not based on a quantitative consideration of the depth of vibration.

The depth of vibration is a critical variable for low-dimensional models that do not include fixed anterior, posterior, or lateral boundary conditions. An effective estimate of the mass in vibration can be made from the current data. The anatomical depth, D_a , at any given location along the anterior-posterior direction for a curved vocal fold (Fig. 11) can be represented using the equation

$$D_a = D_{ap} - ky^2, \quad (4)$$

where D_{ap} is the anatomical depth at the posterior boundary, k is a constant determining the extent of curvature, and y is the location along the anterior-posterior direction. y^2 in the equation comes from the curved vocal fold boundary. Using the boundary condition at the anterior margin

$$D_{aa} = D_{ap} - kL^2 \quad (5)$$

where D_{aa} is the anatomical depth at the anterior boundary, and L is the length of vocal folds (also the distance of the anterior boundary from the posterior end). Rearranging the terms yields the constant

$$k = (D_{ap} - D_{aa})/L^2. \quad (6)$$

The effective mass M of a one-mass vocal fold model with rectangular geometry is (Titze and Story, 2002)

$$M = \rho L T D_{eff}. \quad (7)$$

Here, ρ is the tissue density and T is the thickness of the vocal folds. The effective depth, D_{eff} can be computed from the anatomical depth, D_a as follows:

FIG. 11. Right vocal fold with lateral curved boundary.

$$D_{eff} = c_1 \frac{1}{L} \int_0^L \left[D_{ap} - (D_{ap} - D_{aa}) \frac{y^2}{L^2} \right] dy + c_2. \quad (8)$$

Solving the integration gives

$$D_{eff} = c_1 \left[\frac{1}{3} D_{aa} + \frac{2}{3} D_{ap} \right] + c_2 \text{ cm}. \quad (9)$$

The constants c_1 and c_2 can be obtained from the data generated in the current study. As an example, we computed the constants for the effective depth at the center (row 4) along the superior-inferior direction for the test cases shown in Fig. 8. The constants obtained were listed in Table II and vary slightly as a function of lung pressure.

As an example, the effective mass of a vocal fold with $D_{aa} = 0.2$ cm, $D_{ap} = 1.5$ cm, $L = 1.6$ cm, $T = 0.8$ cm, and $\rho = 1040$ kg/m³ is 0.37 g at a lung pressure of 1.0 kPa. For comparison, the mass used by Flanagan and Landgraf (1968) for a one-mass model was 0.24 g at 0.8 kPa pressure. In the two-mass model of Ishizaka and Flanagan (1972), a lower mass of 0.025 g and an upper mass of 0.125 g were chosen at a similar pressure. In both of these low-dimensional mass-spring models, the masses decreased with the square root of the natural frequency of the mass-spring system. This study has shown how effective mass can be adjusted more realistically with lung pressure and intrinsic muscle activations.

TABLE II. Constants c_1 and c_2 in Eq. (9) obtained at different lung pressures for test cases shown in Fig. 8.

P_L (kPa)	c_1	c_2
1.0	3.636	-3.6017
1.5	3.9268	-3.9263
2.0	2.575	-2.4899
2.5	3.6386	-3.6235

It should also be noted that the conclusions from the current study were dependent on the accuracy of the Fiber-Gel Finite Element model. The results should be further validated by laboratory experiments with modern imaging techniques. Optical coherence tomography is now feasible to a depth of a few mm below the surface of tissues, which may produce a portion of the skin-depth curve (Coughlan *et al.*, 2016). High-speed videoendoscopy can also be used to observe the effective depth of vibration on the superior surface (Zacharias *et al.*, 2018). In addition, the modeling accuracy can be improved with better viscoelastic constants for all tissue layers across vocal fold configurations and muscle activations.

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