Intraoperative Measurement of the Elastic Modulus of the Vocal Fold. Part 1.
Device Development

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Although the concept of manipulating laryngeal biomechanics to improve vocal function is not new, there has been a recent proliferation in surgical techniques used to affect laryngeal function. These include methods which increase the stiffness of the vocal folds, medialize the vocal folds, alter the pitch by changing the tension of the vocal folds, and augment the tissues using injection of alloplastic materials. Despite these new and possibly revolutionary methods, no means are presently available to surgeons to intraoperatively evaluate and optimize results of a surgical intervention. This study involved the development of a device to measure the in vivo elastic modulus of the vocal folds.

INTRODUCTION

The purpose of this ongoing work is to test the hypothesis that the intraoperative measurement of the transverse (horizontal) Young's modulus of the human vocal folds will improve phonosurgical results for glottal incompetence caused by unilateral vocal fold paralysis. A first step toward the examination of this hypothesis included the development of a device to:

1. Measure the vocal fold's elastic modulus rapidly and easily by deflecting the vocal fold a small distance with a measured force applied to a small area of the fold.

2. Provide sensitivity to small changes in the modulus, allowing measurement of differences as small as 5000 dynes/cm^2.

Although the concept of manipulating laryngeal biomechanics to improve vocal function is not new,[1,2] there has been a recent proliferation of surgical techniques to modify laryngeal function.[3–15] These include methods that increase the stiffness of the vocal cords, medialize the vocal cords, raise the pitch of the voice by increasing the tension of the vocal folds, or augment the tissues using injection of alloplastic materials. Despite these new methods, no objective means are presently used by surgeons to evaluate and optimize results of their surgical intervention intraoperatively. Instead, the surgeon usually gauges the changes made to the vocal folds by visual inspection or by listening to the voice and making some perceptual judgment. Improvement is said to occur when a better voice results from adequate augmentation or medialization. In contrast, a poor outcome usually results from miscalculation by the surgeon of the appropriate amount of augmentation, vocal fold medialization, or stiffening. Although there is no published literature concerning success rates for phonosurgery of a large group of patients, clinical experience suggests that a continuum of success exists. There are a number of patients whose surgical results are less successful than desired. A method to improve the success rate would have broad application over a wide range of phonosurgical procedures.

Young's (elastic) modulus was selected as the single measure of choice because its potential for optimizing the outcome of phonosurgery was judged as the greatest among other candidate measures such as acoustic, glottographic, videostroboscopic, or aerodynamic methods. Young's modulus is calculated from the following equation: Y = (Force required for lateral movement/Area over which the force is applied)/(Change in length in X at a particular X), or Stress/Strain.[16]

Physiologically, the elastic modulus of the vocal folds determines, in great part, the wave motion of the mucosal membrane.[17–20] This wave travels from the inferior portion of the folds to the superior, and then moves laterally onto the surface of the vocal folds. In so doing, this wave motion is the primary influence on the rise and fall in the amount of air flowing through the glottis during voicing. The shape of the airflow or volume velocity signal results in the acoustic signal,

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which in turn is used in the perception of voice, *i.e.*, fundamental frequency and associated harmonics of varying amplitudes.[21] Thus, the modulus of the vocal folds may be an important factor affecting the sound produced by the larynx.[22,23]

Less than 15 years ago, Baer[17] quantitatively described for the first time the wave nature that had been observed by Timcke, *et al.*[24,25] based upon high-speed film of the vibrating larynx. Baer noted that a mucosal wave travels from the inferior portion of the vocal folds to the superior and then breaks onto the surface of the folds like a wave on the shoreline. Baer determined that it traveled at approximately 2 m per second. He has since theorized that the speed of the wave may not be constant. Variation in wave speed may be one of the determining factors in pitch control according to the relationship $\text{Freq} = k(\text{wave speed}/\text{wavelength})$, where k is a constant or function of proportionality.[26]

As a corollary to Baer's work, a number of authors have shown that loss of this traveling wave is important in understanding dysphonia. Loss of the wave in paralysis has been associated with diminution in the amplitude of some of the harmonics in the acoustic signal.[27,28] Loss of the wave in vocal fold scarring and/or neoplasia has also been associated with the appearance of energy at inharmonic frequencies, *i.e.*, noise in the source spectrum.[29,30] Not only is the wave associated with the acoustics of the source signal, but recent work on the mechanics of vocal fold vibration has shown that the traveling wave may be the fundamental element of vocal fold vibration.[31]

A number of factors are pertinent to the formation and characteristics of the traveling wave. Hirano[32] proposed the body cover theory of vocal fold vibration. It suggests that the layer structure of the vocal fold divides into two groups which have different rheological properties. The cover is composed of squamous epithelium and the superficial and intermediate layers of loose connective tissue or lamina propria. Thus, it is very pliable but has no intrinsic contractile properties. Conversely, the inner group or body is composed of the deep layer of the lamina propria and the thyroarytenoid muscle. Cricothyroid muscle contraction produces stretching of the cover and thyroarytenoid muscle contraction produces stiffening of the body. The combined stiffness of the entire vocal fold would then be determined by extrinsic longitudinal tension on the cover and the internal stiffening of the body. In this theory, vocal fold vibration occurs primarily in the cover.[7,33]

In an effort to explain why the larynx oscillates and delineate the factors pertinent to various vibratory modes, a number of authors have developed theoretical mathematical models of vocal fold motion as an approximation of the physiologic traveling wave. Presently, the "gold standard" by which most laryngeal models are evaluated is the two-mass model, developed by Ishizaka and Flanagan in 1972,[34] based upon a theoretical treatise by Ishizaka and Matsudaira.[35] In this computer model, laryngeal vibration is concisely modeled using two masses representing the lower and upper vocal fold margins and connecting springs. Most models of vocal fold function require the determination of stiffness or mechanical compliance parameters in order to define the motion of the masses constituting the models.[19,34–36] Because no in vivo experimental values for Young's modulus have previously been published, a number of authors have measured the elastic modulus constants of the larynx in vitro to predict how laryngeal vibration may actually work in the in vivo state.

One of the first in vitro measurements was by Ishizaka and Kaneko,[37] who removed a freshly cut human larynx, fastened a thread to a vocal fold, and stretched the vocal cord laterally to measure the displacement. From this preparation, they determined the stiffness constant K_1 which was stated to be 37×10^3 dynes/cm. Perlman and Durham[38] performed in vitro studies on the vocal fold mucosa during isometric contraction, and developed force elongation curves and stress-strain curves for vocal fold mucosal specimens. They calculated mean longitudinal secant modulus for the vocal fold cover ranging from 1.1×10^6 dynes/cm^2 at 10% strain to 1.8×10^6 dynes/cm^2 at 50% strain. They also determined an overall value for Young's modulus for the total cover and vocalis muscle which ranged from about 1×10^6 dynes/cm^2 at 10% strain to 4×10^6 dynes/cm^2 at 50% strain. In another study, Kakita, *et al.*[39] measured Young's modulus of the vocal folds in isotropic longitudinal planes as well as orthotropic (anisotropic) vertical and horizontal planes. Stevens[40] used Kakita's values for Young's modulus to calculate the motion of the lower margin in his laryngeal model. The proposed vocal fold motion is based on an equation which uses the inverse of the modulus or compliance, fold thickness, fold mass, and subglottal pressure. His model uses a value of 3.3×10^4 dynes/cm^2 for the modulus of the vocal fold.

The study of Young's modulus in pathologic states of the larynx holds promise for illuminating underlying pathophysiologic mechanisms. For example, Trapp, *et al.*[23] studied states of simulated laryngeal paralysis in a canine model. Teflon® injection of the paralyzed vocal fold to the point where the mechanical compliance ($1/Y$) or stiffness between the two vocal folds became roughly equivalent was associated with return of the traveling wave and a subsequent increase in the amplitude of the upper harmonics of the source signal. This study indicated that not only is the elastic modulus important in determining the nature of normative vocal fold vibration, but that asymmetry in the elastic modulus between the vocal folds may have a profound effect on abnormal vocal fold vibration. In fact, it was found that even small differences in the stiffness between the two vocal folds, produced

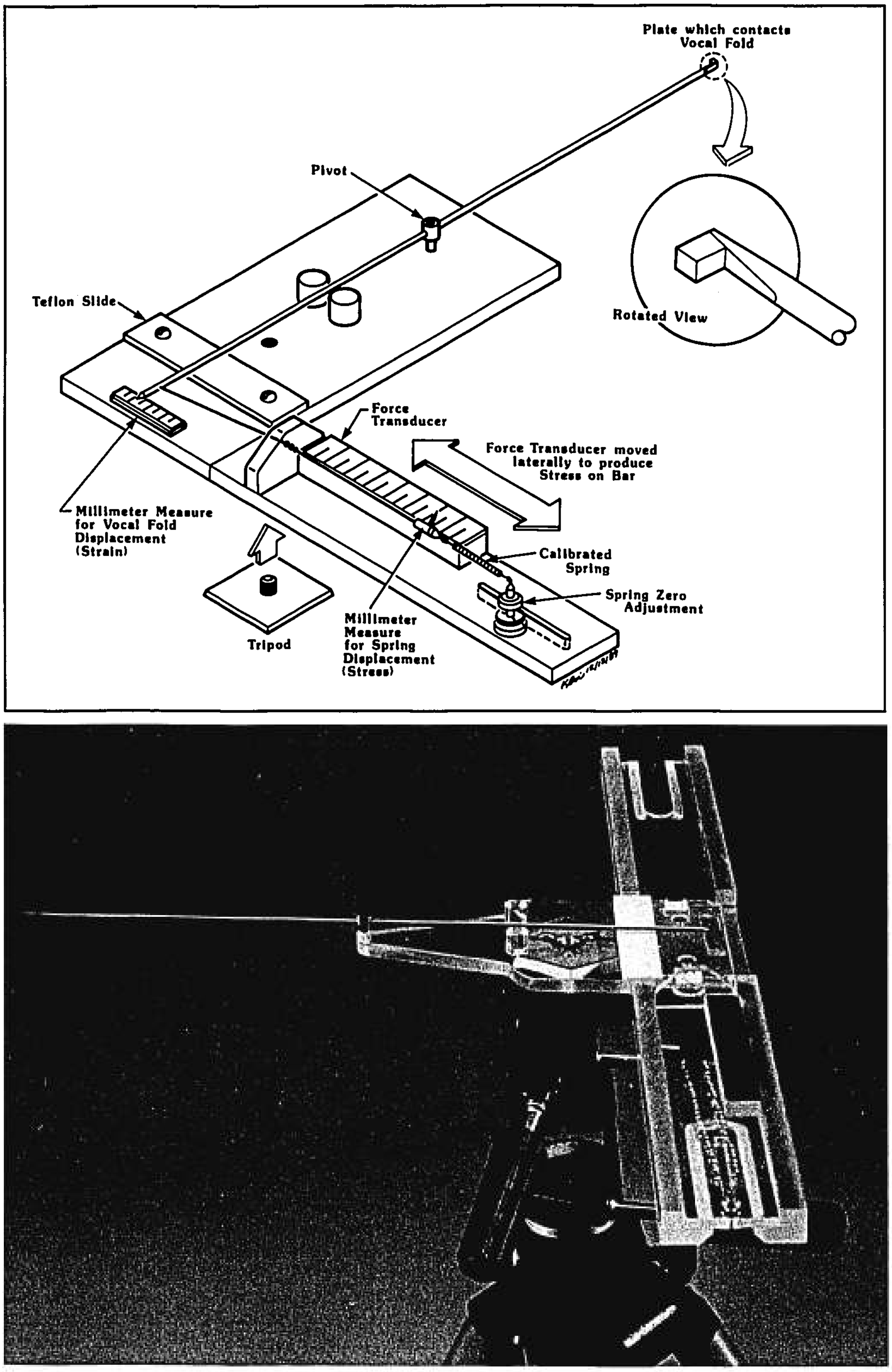


Fig. 1. **Top.** Schematic of the device to measure Young's modulus. **Bottom.** Photograph of revised human prototype device to measure Young's modulus.

by less than a 0.1-mA stimulation difference between the recurrent laryngeal nerves in the in vivo canine model, produced phonation with irregular vibration and diplophonia.[41]

The belief that surgeons would someday benefit by an ability to objectively determine the vocal fold's elastic properties prompted development of a simple device to measure Young's modulus in vivo.

METHODS

Device Construction

A prototype device was constructed to measure Young's modulus in vivo. The device was constructed with a large vocal fold contact area (40%) to improve sensitivity for in vivo canine experimentation. It was later modified to a reduced contact area of 10% of the vocal fold area (6 mm^2) in order to approach more classical determinations of Young's modulus.[42] A diagram and photograph of the prototype model is presented in Figure 1. The device consists of a platform on which a metal bar pivots so that the bar deflection is measured. A small foot at one end of the bar is applied to the vocal fold. The force on the bar required to produce a lateral fold deflection is calculated using a calibrated spring, using the equation, Force = (spring constant) × (distance the spring is stretched). Initial displacement is set by first increasing the force on the bar to produce a 50% strain (one-half maximal displacement or 2 mm, given the vocal fold thickness in the medial-lateral dimension of approximately 4 mm). The force on the bar is then released so that it returns to a resting position. The resulting position is designated as the initial zero displacement.

Theoretical Concerns

A number of theoretical concerns have been addressed. One concern was the direction of the modulus to measure. One could have measured the longitudinal, vertical, or transverse elastic moduli.[18,39,43,44] The longitudinal elastic modulus in the anteroposterior direction is the most isotropic; however, it was not possible to measure it in an intact larynx. Furthermore, longitudinal tension effects were probably included in measurement of transverse stress-strain curves. The vertical modulus is probably different than the transverse. However, the vertical motion of the vocal fold is quite small when compared to the transverse component of motion, and vertical stress can only be applied from below the glottis. A number of studies have determined that when the vocal fold vibrates, a puff or bubble of air starts at the inferior margin and travels superiorly, causing a mucosal wave to travel from inferior to superior.[45,46] However, while the direction of the wave motion is inferior to superior, this is accomplished by vocal fold displacement in a transverse direction. For the aforementioned reasons, it was believed that the transverse (horizontal) modulus was the most important in determining the biorheology of vocal fold motion.

An additional concern, given the viscoelastic nature of vocal fold tissues, was that significant dispersive effects could occur. It is quite possible that as the traveling wave reaches the superior margin of the vocal fold and spreads out laterally, dispersive effects significantly change the wave speed, because the folds are not purely elastic. However, it was presumed that this phenomenon was always present in roughly the same proportion for various vibration modes and that viscous tissue losses were small when compared to tissue elastic forces.

Another consideration is that relying on one measurement to define the fold's elastic characteristics implies a one-dimensional elastic body, which is known to be incorrect. However, many of the features of extremely complicated orthotropic elastic systems such as in vascular hemodynamics[47] and pulmonary physiology[48] are accurately described by one-dimensional approximations.

Finally, the goal was to develop a device to measure a single parameter which could be easily used by a physician as a guide during surgery, rather than all parameters pertinent to vocal fold motion.

Practical Concerns

Development of the device required that four prerequisites be met. First, it was important that the device measure Young's modulus rapidly and easily. This necessitated that the force to move the bar be rapidly applied, be stable, and be easily read, because of strain "creep" and stress "relaxation" in vocal fold tissue.[39] Creep and relaxation are factors that must be considered in designing and using a device to measure Young's modulus. Stress relaxation was observed in the preliminary studies when force was applied for longer than 30 seconds.

Second, the device had to be sensitive to small changes in the modulus. Sensitivity of the prototype model appeared to be determined, to a large degree, by the ability to quickly read the changes in bar displacement. Initially, a calibrated spring was selected with a sensitivity of 5000 dynes per 0.25-cm displacement.

Third, it was necessary that the device be manuevered into position easily so measurements could be made at frequent intervals. In order to dynamically measure the elastic modulus, it was required that the device either be able to remain within the introitus of the larynx while surgery was being performed or be easily taken in and out of the laryngoscope during frequent sequential measurements. For example, Teflon is injected lateral to the vocalis; thus, the device's midline deflection bar does not interfere with injector-needle placement through the laryngoscope. The bar itself had to take up very little room in order to permit concomitant laryngeal microsurgery. The entire device also had to be small enough to be easily maneuvered by the surgeon.

Finally, the device had to improve the surgical outcome for glottal incompetence. For the device to significantly improve patient treatment, it must be demonstrated that, by using the device, physicians can more readily assure themselves of a good surgical outcome. This aim continues to be the subject of ongoing study.

In Vivo Canine Model

The device used for the in vivo canine model is similar to that depicted in Figure 1 except that it has a larger area of vocal fold contact to increase sensitivity. A small metal plate (approximately 0.7 × 0.15 cm) at one end of the bar is applied to the vocal fold. The area of contact is 0.7 × (vertical height of the vocal fold).

Figure 2 demonstrates the in vivo canine model in which the device was initially developed and tested. Five mongrel dogs (25 kg) were premedicated with Innovar® intramuscularly. Intravenous Pentothal® was administered to a level of corneal anesthesia, and additional Pentothal was used to maintain this level of anesthesia throughout the experiment.

The animal was placed supine on the operating table, and a midline incision was made to expose the trachea from the hyoid to the sternal notch. Both recurrent laryngeal nerves were identified and preserved. Both superior laryngeal nerves were identified along their course to the cri-

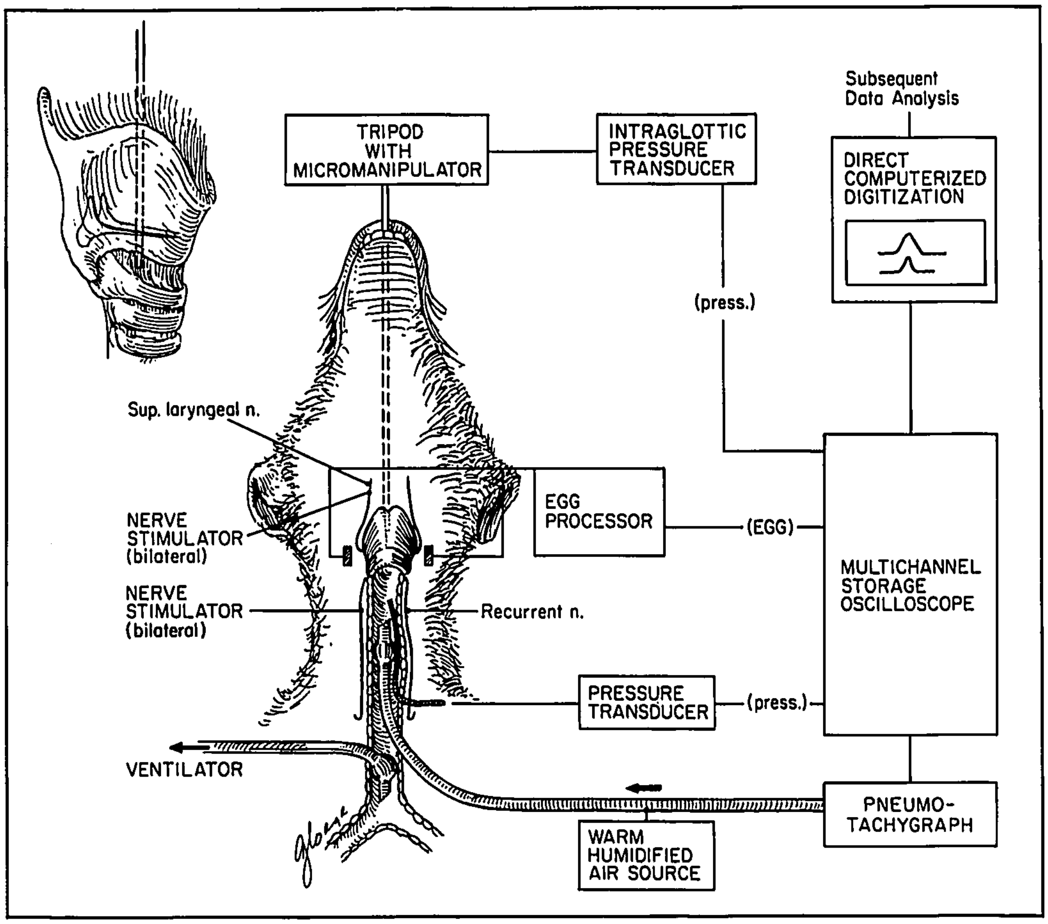


Fig. 2. Diagram of in vivo canine preparation.

cothyroid muscles. A low tracheotomy was performed at the level of the suprasternal notch, through which an endotracheal tube was passed to allow ventilator assisted respirations. A second tracheotomy was performed in a more superior location, through which a cuffed endotracheal tube was passed in a rostral direction and positioned with the tip 10 cm below the vocal folds. The cuff was inflated to just seal the trachea. Humidified heated air was passed through this rostral endotracheal tube from a compressed air tank. Flow was controlled with a valve and measured with a Gilmont flowmeter (Great Neck, NY: Model 4) and a pneumotachograph (OEM Fleisch #7, Richmond, Va.). The pneumotachograph was then used with a differential pressure transducer (Fluid Precision Inc., Billerica, Mass. Model #183) to record input airflow. The airflow was humidified and heated by bubbling it through 5 cm of heated water so that the temperature of the air was 37°C when measured at the glottic outlet.

One centimeter segments of recurrent and superior laryngeal nerves were isolated, and Harvard miniature electrodes were applied around each nerve. The electrodes were then insulated from surrounding tissue. A constant current nerve stimulator (WR Medical Electronics Co., St. Paul, Minn.: Model S2LH) was used to stimulate the recurrent laryngeal nerves (RLN) and a constant voltage source (Grass, Quincy, Mass., Model 54H; WPI, New Haven, Conn., 301-T) was used to stimulate the superior laryngeal nerves (SLN). These nerves were stimulated at 70 to 80 Hz, with 0.3 to 0.5 mA (RLN) or 1 to 1.4 V (SLN) intensity for a 1.5-msec pulse duration. There was no observed contraction of the cricothyroid during maximal stimulation of the RLN at 2.0 mA. Similarly, there was no bulging or contraction of the thyroarytenoid muscle and no movement of the arytenoid during maximal stimulation of the SLN at 3.0 V. Phonation was produced with an airflow of 318 cc/s applied through the larynx by the rostral endotracheal tube.

Upstream subglottic pressure was measured using a Millar (Houston, Tex.) Mikro-Tip catheter pressure transducer (Model No. SPC-330, Size 3F). The subglottic pressure transducer was passed rostrally through the superior tracheotomy and placed 5 cm below the glottis.

Electroglottographic (EGG) signals were obtained with a Laryngograph (Synchrovoice, Harrison, NJ) with the two recording electrodes sutured into place on the right and left thyroid ala, just above the cricothyroid muscles. The reference electrode was secured to an adjacent strap muscle. The EGG, input airflow, and supraglottic and subglottic pressure signals were low-pass filtered at 1 kHz, digitized at 2.5 kHz for 20 seconds, and stored on the hard disk of a personal computer.

Stroboscopic videoendoscopy was performed using a B&K 4914 stroboscope to illuminate the glottis through a zero-degree documentation sheath, connected to a Jed Med CCD Camera (St. Louis, Mo.: Model 70-5150) and recorded on a SONY video recorder (Park Ridge, NJ: Model VO5850).

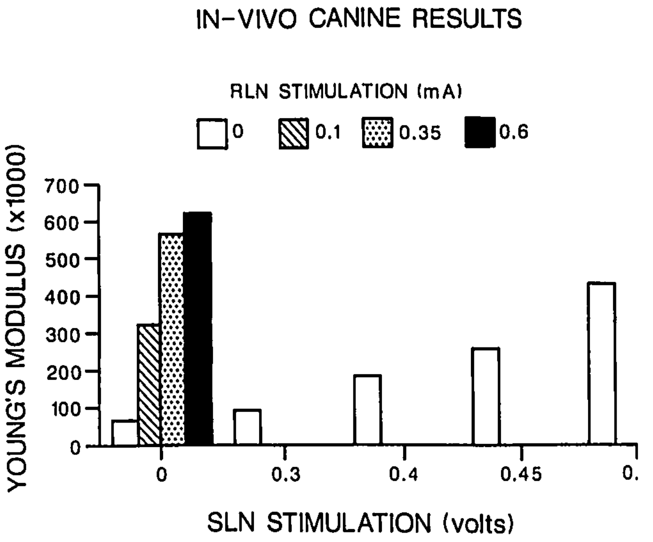


Fig. 3. Young's modulus values for recurrent (RLN) and superior (SLN) stimulation.

RESULTS

In Vivo Canine Data

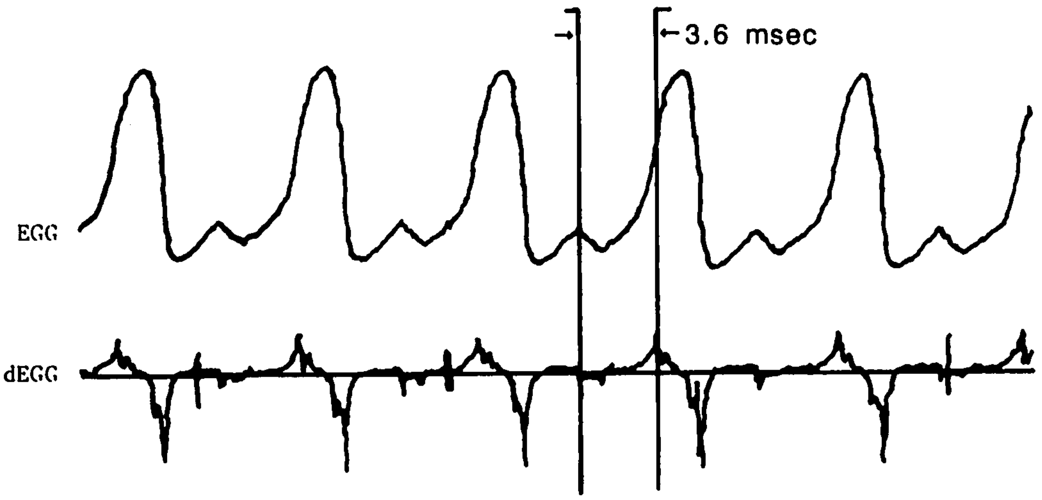
Using the device, in vivo measurements of Young's modulus were derived for the canine's vocal fold at rest, with superior laryngeal nerve (SLN) stimulation, with recurrent laryngeal nerve (RLN) stimulation, and for combined recurrent and superior laryngeal nerve stimulation. Figure 3 demonstrates that the device responded to changes in both intrinsic stiffness (RLN stimulation) and extrinsic tension (SLN stimulation).

In addition, values for the velocity of the traveling wave (based on *Y*) to measured values from electroglottographic traces were compared. Figure 4 demonstrates this comparison. The upper part of Figure 4 depicts an EGG trace. The cursors depict the time interval from lower margin separation to upper margin separation (3.6 msec). The measured distance between the lower and upper margins was 0.7 cm (over the twice the value in humans). This gives a measured velocity of 194 cm/s. The lower half of the figure demonstrates the calculation of wave speed by using the relationship of Wavespeed = (Young's modulus/tissue density)$^{1/2}$. This equation was chosen because the speed of a wave in an elastic medium can be determined by the square root of a term representing the resistance to deformation *(Y)* divided by a term representing the inertia of the medium (density).[43] Using this equation it can be seen that, as the stiffness or elastic modulus of the vocal fold increases, so would the traveling wavespeed and therefore the pitch. The differential Young's modulus was chosen as suggested by Kakita, *et al.*[39] and Stevens.[40] The force used to displace the 0.56 cm^2 contact plate 0.1 cm was 20,053 dynes. Therefore, the stress = 20,053 dynes/0.56 cm^2 or 35,810 dynes/cm^2. A maximum transverse displacement for cord movement was estimated at 0.1 cm during vibration at the input airflow of 300 cc/s. This value was obtained from measurements taken from videostroboscopic images at a flow rate of 300 cc/s.

The differential strain is the change in displacement over the initial displacement or 0.1 cm/0.1 cm, or 1.[39] Using these values of stress and strain gives a differential Young's modulus of 35,810 dynes/cm^2 for the right cord. In a similar fashion the differential modulus for the left cord was 51,477 dynes/cm^2. The average modulus for both folds was 43,644 dynes/cm^2. In order to calculate a wavespeed, tissue density must be considered. Given the tissue density of the lamina propria and epithelium as 1 gm/cm^3, the calculated wave speed is (43,644/1)$^{1/2}$ or 209 cm/s. This value of 209 cm/s for the calculated wavespeed is quite close to the measured value of 194 cm/s.

Figure 5 demonstrates a similar comparison for a higher level of nervous stimulation at a flow rate of 500 cc/s. Again the measured value of 292 cm/s compares favorably to the calculated value of 288 cm/s. Although certain assumptions were required for these comparisons, the data demonstrate that increasing stimulation to the recurrent laryngeal nerves was associated with a higher fundamental frequency and an increase in the value of Young's modulus.

Preliminary tests of the device's utility in the canine model were performed by measuring the secant Young's modulus of the vocal folds, the traditional method for reporting Young's modulus. $Y = 123{,}000$ dynes/cm^2 at 50% maximum strain when a 1.35-mA current was delivered to each recurrent laryngeal nerve and 0.56 V to each superior laryngeal nerve. The resulting glottographic waveforms are shown in Figure 6. Normal simulated phonation was produced by changing recurrent laryngeal nerve stimulation to maximize vocal efficiency, the ratio of the output sound power at the mouth to input aerodynamic power.[49] The level of stimulation associated with maximum vocal efficiency was approximately equal to the average of two stimulating currents: one of the currents set to produce initial medial arytenoid movement and the other set to produce maximal arytenoid movement with vocalis bulging. Paralysis was then simulated by cutting the left recurrent laryngeal nerve. The modulus of the paralyzed fold was found to be $Y = 85{,}000$ dynes/cm^2. Vocal efficiency fell to about 10% of simulated normal efficiency. The traveling wave was reproduced by injecting Teflon into the paralyzed fold until the modulus of the paralyzed fold approximated that of the normal fold. Vocal efficiency rose to approximately 40% of the simulated normal efficiency. Injection of 0.2 cc of Teflon produced a modulus in the paralyzed fold of 119,000 dynes/cm^2. The prototype device was able to measure the effect of small amounts of augmentation on the fold's modulus and could have assisted in determining an endpoint for vocal fold augmentation. Interestingly, intentional overinjection of Teflon to 0.4 cc produced loss of the traveling wave and a modulus of 147,000 dynes/cm^2.



Measured wave speed = C_{mea} = (Distance)/(Time) = (cm/sec)

Measured: 3.85, 3.45, 3.5 msec Average = 3.6 msec = 0.0036 sec
Stimulation: 0.125 mAmp to RLNS; 0.7 Volts to SLNS
Distance from lower to upper margin = 0.7 cm
Frequency = 122 Hz Input Air Flow = 300 cc/sec
Distance = 0.7 cm, Time = 0.0036 sec:
Therefore, C_{mea} = (0.7 cm)/(0.0036 sec) ≐ **194 cm/sec**

Right vocal cord					**Left vocal cord**				
Spring Displacement		Vocal Cord Displace.			Spring Displacement		Vocal Cord Displace		
Initial disp. = 1.2 cm					Initial disp. = 1.25 cm				
Data Xs	ΔXs	Xvc	Strain	ΔYoung's	Data Xs	ΔXs	Xvc	Strain	ΔYoung's
2.0 cm	0.8 cm*	0.1 cm	1.0	35810**	2.4 cm	1.15 cm	0.1 cm	1.0	51477
2.75	0.75	0.2	0.5	67144	3.0	0.6	0.2	0.5	53715
3.6	0.85	0.3			2.8		0.3		

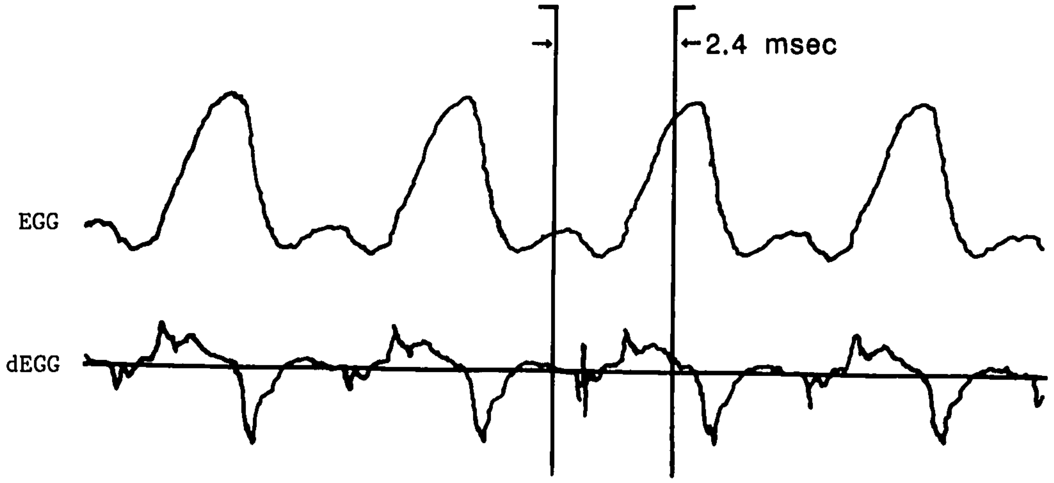
Calculated Wave Speed = C_{calc} = (ΔY modulus/density of tissue)$^{1/2}$
Hooke's Law: F = kx (F = force, k = spring constant, x = displacement); k = 25067 dynes/cm
Vocal Fold Plate Contact Area = 0.56 cm^2
Maximum Displacement of vocal cord = 0.1 cm at air flow = 300 cc/sec
Density of air = 1 gm/cm^3

$\Delta Xs_i = Xs_i - Xs_{i-1}$ $\Delta Xvc_i = Xvc_i - Xvc_{i-1}$
*ΔXs = 2.0 - 1.2 = 0.8 cm
** Differential Young's modulus = Stress/Strain = 35810/1
Strain = ΔXvc/Xvc = 0.1/0.1 = 1 Stress = ΔForce/Area = (k x ΔXs)/0.56 = (0.8 x 25067)/0.56 = 35810
Average Differentiated Young's Modulus = 35810 + 51477 = 43644
Therefore C_{calc} = $(43644/1)^{1/2}$ = **209 cm/sec**

Fig. 4. Comparison of measured traveling wave speed to calculated traveling wave speed. EGG = electroglottograph; dEGG = differentiated electroglottograph.

DISCUSSION

Sensitivity of the prototype model appeared to be determined to a large degree by the ability to quickly read the changes in bar displacement. To increase this ability, it may be necessary to lessen the parallax error between bar movement and the underlying millimeter measuring stick. Another solution may be to offset the pivot point so that the pivot is closer to the plate end of the bar, thereby increasing the excursion of the measuring end of the bar. This would require an alteration in the force calculation due to the mechanical advantage of moving the pivot point. Another issue is the ability to apply small forces and keep them constant during the reading of displacement. Initially, a calibrated spring was chosen with a sensitivity of 5000 dynes per 0.25-cm displacement. The



Measured wave speed = C_{mea} = (Distance)/(Time) = (cm/sec)

Stimulation: 0.150 mAmp to RLNS; 0.7 Volts to SLNS
Frequency = 210 Hz Input Air Flow = 500 cc/sec
Distance from lower to upper margin = 0.7 cm, Time = 0.0024 sec
Therefore, C_{mea} = (0.7 cm)/(0.0024 sec) = **292 cm/sec**

Right vocal cord					**Left vocal cord**				
Spring Displacement		Vocal Cord Displace.			Spring Displacement		Vocal Cord Displace		
Initial disp. = 0.4 cm					Initial disp. = 0.4 cm				
Data Xs	ΔXs	Xvc	Strain	ΔYoung's	Data Xs	ΔXs	Xvc	Strain	ΔYoung's
1.3 cm	0.9 cm	0.1 cm	1.0	40286	1.3 cm	0.9 cm	0.1 cm	1.0	40286
2.25	.95*	0.2	0.5	85049**	2.2	0.9	0.2	0.5	80572
3.8		0.3			2.8		0.3		

Calculated Wave Speed = C_{calc} = (ΔY modulus/density of tissue)$^{1/2}$
Hooke's Law: F = kx (F = force, k = spring constant, x = displacement); k = 25067 dynes/cm
Vocal Fold Plate Contact Area = 0.56 cm^2
Maximum Displacement of vocal cord = 0.2 cm at 500 cc/sec input air flow
Density of air = 1 gm/cm^3

$\Delta Xs_i = Xs_i - Xs_{i-1}$ $\quad \Delta Xvc_i = Xvc_i - Xvc_{i-1}$
*ΔXs=2.25 - 1.3 = 0.95
** Differential Young's modulus = Stress/Strain = 42524/0.5 = 85049
Strain = ΔXvc/Xvc = 0.1/0.2=0.5
Stress = ΔForce/Area = (k x ΔXs)/0.56 = (25067 x 0.95)/(0.56) = 42524
Average Differentiated Young's Modulus = 85049 + 80572 = 82811
Therefore C_{calc} = (82811/1)$^{1/2}$ = **288 cm/sec**

Fig. 5. Comparison of measured traveling wave speed to calculated traveling wave speed. EGG = electroglottograph; dEGG = differentiated electroglottograph.

next revision of the device will substitute a vernier force transducer for the calibrated spring currently in use.

In the next phase of this research, the laryngeal nerves will be transtracheally stimulated to arrive at the elastic modulus during simulated phonatory posture. There are a number of reports in the literature of safe stimulation of the recurrent laryngeal nerves intraoperatively during thyroid surgery to assist in identification of the nerves.[50,51] The nerve is positively identified when electrical stimulation produces a palpable movement of the vocal fold beneath the cricothyroid membrane. If a rostral airflow is provided through the larynx, phonation can be produced by direct electrical stimulation of the recurrent laryngeal nerves,[52,53] although this requires major surgery.

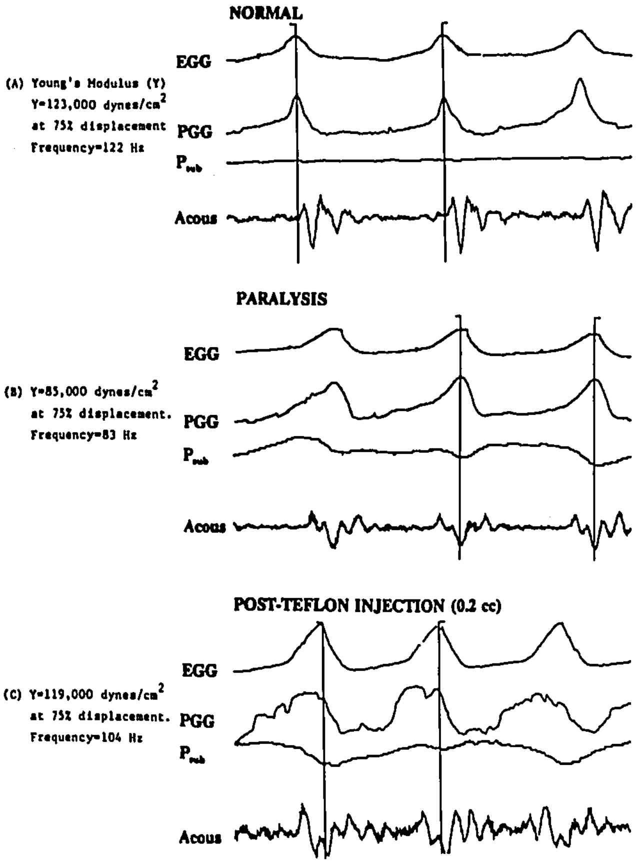


Fig. 6. Electroglottographic (EGG), photoglottographic (PGG), subglottic pressure (Psub), and acoustic signals for simulated normal **(A)**, paralysis **(B)**, and post-Teflon injection **(C)** phonation states.

The nerves to the muscles of the larynx were electrically stimulated to produce phonation during general anesthesia in canine preparations by a transtracheal stimulation of the recurrent laryngeal nerves.[54] Several authors have safely used transcutaneous and transesophageal techniques for stimulation of the recurrent laryngeal nerves in humans.[55,56] During the next phase of this study, transtracheal stimulation of the unaffected vocal fold will be performed to arrive at the appropriate target modulus for Teflon augmentation.

CONCLUSIONS

1. Young's modulus is an important variable which defines, in great measure, the biorheology of the vocal fold's traveling wave.

2. The elastic modulus of the vocal fold is determined by both the stiffening of the body and the tension of the cover.

3. Acute laryngeal paralysis and paresis are associated with a decrease in the elastic modulus of the vocal fold.

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Second Asian Congress of Oral Surgery Set

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