A novel ultrasound technique to study the biomechanics of the human esophagus in vivo

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THE BIOMECHANICAL PROPERTIES of the wall of the esophagus are important determinants of the motor and sensory function of the esophagus. Both active (muscle contractions) and passive (elastic and visco-elastic properties) properties of the esophageal wall contribute to these biomechanical properties. The biomechanical properties of a tissue are expressed through its stress-strain relationship. A number of investigators have attempted to determine the circumferential tension-strain or stress-strain relationship of the esophagus by using various in vivo and in vitro techniques (1, 2, 4, 5, 8). To determine such a relationship, one must measure both the loading (pressure) and the deformation [changes in cross-sectional area (CSA) and wall thickness] simultaneously. To date, there has been no method to measure these parameters simultaneously in the human esophagus in the in vivo state. Gregersen et al. (5) developed the technique of impedance planimetry for measurement of pressure and CSA of the gastrointestinal tract. Their technique, however, does not measure the wall thickness and therefore allows computation of wall tension (pressure × radius) but not wall stress (pressure × radius/wall thickness). Jorgensen et al. (8) combined the impedance planimetry method with A-mode ultrasound to measure the wall thickness during distension. However, wall thickness was measured in only one direction and CSA was calculated indirectly on the basis of the impedance of the saline solution inside the balloon.

The purpose of the present study is to introduce a rather simple, novel technique that combines manometry with B-mode ultrasonography to measure pressure, radius, and wall thickness of the esophagus directly, simultaneously, and continuously. This novel technique allows calculations of the circumferential stress, strain, and elastic modulus on a continuous-time basis. We performed in vitro studies to validate the accuracy of the technique and then studied the stress-strain relationship of the esophageal wall in healthy normal subjects.

MATERIALS AND METHODS

In vitro studies. An in vitro validation study was performed using a 20-MHz, high-frequency intraluminal ultrasonography (HFIUS) probe (UM-3R; Olympus, Tokyo, Japan) positioned in water in the agar tubes. Three tubes of different wall thickness and lumen size were constructed from 3% agarose. Measurements of the wall thickness and inner radius were made using a caliper and the ultrasonographic method with A-mode ultrasound to measure the wall thickness during distension. However, wall thickness was measured in only one direction and CSA was calculated indirectly on the basis of the impedance of the saline solution inside the balloon.

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the center of the tube, off center, and center of the tube with a 15° angle between the transducer and the wall of the tube. A composite tube with an inner layer of 1% agarose and an outer layer of 3% agarose was also constructed to simulate a multilayered esophagus. For these measurements, the HFIUS probe was placed vertically in the center of the tube.

**In vivo studies.** The study was performed in 18 healthy volunteers, 13 men and 5 women. The age of the volunteers ranged from 18 to 62 years (median 32.7 years). Volunteers with a history of upper gastrointestinal surgery or systemic diseases known to influence gastrointestinal motility and volunteers that were taking medications that can affect the esophagus were excluded from the study. The study was approved by The Human Subjects Committee of the University of California, San Diego.

**Measurement system.** An esophageal catheter assembly consisting of a 6.2-Fr catheter equipped with a HFIUS probe, a manometry catheter, and a high-compliance polyvinyl bag was used for the in vivo studies, as shown in Fig. 1.

The length and maximum diameter of the bag were 5 cm and 3.5 cm, respectively. The distensibility of the bag was measured in vitro, before its use in each subject, to determine the range of bag volume used for in vivo studies. We were able to inflate this 0.03-mm-thick bag to a volume of 25 ml without stretching the bag wall. This maximal volume was never exceeded in the in vivo studies, which ensures that the elastic properties of the bag itself do not contribute to the intrabag pressure in the in vivo situation. The ultrasound probe catheter was anchored at the proximal and distal ends of the bag, which ensured that the ultrasound transducer stayed in the center of the bag.

The manometric catheter assembly had four side holes, with openings located inside the bag and at 2, 7, and 12 cm proximal to the bag. The 2, 7, and 12 cm holes were perfused with distilled water at a rate of 0.5 ml/min using a low-compliance pneumohydraulic perfusion system (Arndorfer Medical Specialties, Greendale, WI). The pressure recordings from the four ports of the manometric catheter were input into a personal computer via a polygraph (Medtronic Synectics Medical, Stockholm, Sweden). The ultrasound and manometric recordings were synchronized using a video timer.

**Study protocol.** After a >5-h fast, the volunteer’s throat was sprayed with lidocaine HCl (Xylocaine Astra, Westborough, MA). With the subject sitting in the upright position, the lubricated assembly was passed transnasally or transorally until the tip was 60 cm from the nostril or 55 cm from the incisors. The subject was then placed in the semirecumbent position. After an adjustment period of 5–10 min, the assembly was pulled back in 1-cm steps while the manometric pressure recordings were observed on the computer monitor. The lower esophageal sphincter (LES) was identified as a high-pressure zone that relaxed in response to a swallow, and its location was noted in relation to the nostril or incisors. The center of the bag was then positioned at 7.5 cm above the LES, and the catheter assembly was anchored at the nostril or at the angle of the mouth with adhesive tape.

After a rest period of 10 min, the bag was inflated with various volumes of water for at least 20 s by injecting water into the bag via a manually held syringe (isovolumic study). The rate of injection was ~2 ml/min. This constituted the short-period distension study. The infused volume was set at 7.5, 12.5, 5, 10, 15, 20, and 17.5 ml, in that order. The steady state point was defined as the time at which intrabag pressure had been constant and the lowest. Each distension was followed by a 30-s rest period. Each volume was tested three times. After this short-period distension protocol, 15 ml distension for 3 min was also performed and repeated two times in each subject. This constituted the sustained esophageal distension study. Esophageal distensions were also performed by using a constant-pressure technique (isobaric study). The bag was distended with water at 20, 40, 60, 10, 30, and 50 mmHg, in that order, via a water reservoir (sterile water bag) placed at the appropriate vertical height. It took ~20–30 s to reach the maximum distension, and the distension was sustained for another 20 s. The steady-state point was defined as that at which the CSA of the lumen reached a maximum. Each distension was followed by a 30-s rest period. Each pressure was tested two times. After this short-period distension protocol, 40 mmHg distension was performed for 3 min and repeated two times in each subject. Ten subjects participated in the isovolumic study and 8 subjects in the isobaric study. During bag distension, the subjects were asked to keep still, to make no attempt to speak, and to refrain from swallowing.

To determine the esophageal dimensions of a relaxed esophagus, we injected atropine (15 μg/kg) through an antecubital vein. An image of the esophagus was captured, at zero bag pressure, ~10 min after the injection.

**Data measurements.** The ultrasound images were recorded in real time using a high-resolution ultrasound unit and a videotape recorder (Sony, Tokyo, Japan). The ultrasound images were digitized on a personal computer equipped with a high-definition video card (Targa; Truevision, Indianapolis, IN) and analyzed by using a commercially available image analysis software package (Mocha; Jandel Scientific, San Rafael, CA). Images were displayed on a 17-inch high-resolution monitor with pixel size of 640 × 480. This corresponds to an image magnification of approximately ×12 (10 pixels = 1 mm). The ultrasound image was captured at a steady state point as mentioned previously. The intrabag pressure was also measured at the same time. The perimeters of the bag and outer esophageal wall were traced for each image by using a computer program. The former corresponds to the inner circumference, and the latter corresponds to the outer circumference of the esophageal wall. Once outlined, the software program automatically calculated the luminal CSA and circumferential length of the esophagus (ℓa). The

![Fig. 1. A schematic diagram of the catheter assembly. Note that soft vinyl support is used to place the high-frequency intraluminal ultrasonography probe at the central axis of the bag.](http://ajpgi.physiology.org/DownloadedFrom/)
Biomechanical analysis. The circumferential deformation of the esophagus may be described by the Green’s strain ($\varepsilon$), which is defined as follows

$$\varepsilon = \left( \frac{1}{2} \left( \frac{L}{L_0} \right)^2 - 1 \right)$$

(1)

where $L_0$ is the circumferential stretch ratio given by $L_0 = l_0/L_0$, where $l_0$ is the circumferential length of the esophagus at a given distension and $L_0$ is the circumferential length at zero pressure after the injection of atropine (15 $\mu$g/kg iv). Because the outer muscle wall was the most prominent and easily recognized on the ultrasound images, the outer wall strain was computed during distensions.

At an equilibrium condition, the average circumferential Kirchhoff’s stress in the esophageal wall at a given distension can be computed according to the following equation, with an assumption that the shape of the esophagus is cylindrical

$$\sigma = \frac{Pr}{h\lambda^2}$$

(2)

where $P$ is the luminal pressure and $r$ and $h$ are the radius and wall thickness of the esophagus, respectively. Ideally, one should relate the mean stress to the midwall strain. However, because the data on the outer circumference is more accurate, we calculated the mean stress and strain at the outer wall. Hence, the depicted data correspond to the biomechanical properties of the outer layer of the esophagus.

Statistical analysis. Some of the data were normally distributed and others were not; therefore, we used the Mann-Whitney rank-sum test for statistical comparisons. The results are shown as means $\pm$ SE.

RESULTS

In vitro studies. There was excellent agreement between the measurements made by the calipers and the ones made by analysis of the ultrasound images. The most accurate measurements of radius of the tube and wall thickness were made when the HF1US probe and the transducer were located in the center and parallel to the wall of the tube. When the transducer was located at an off-center position or at an angle, the entire circumference of the tube was not well visualized. However, measurements at four quadrants could be performed from the portions of the tube that were visualized adequately even during such off-center and angled positions, and it showed that the error was <4.0%. The composite tube was also well visualized with a measurement accuracy of >98% (Table 1).

<table>
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<tr>
<th>Table 1. Differences between actual and ultrasound values</th>
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<tr>
<td>Agar Tubes</td>
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<td>(US value $-$ actual value)/actual value</td>
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<tr>
<td>Tube 1</td>
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<tr>
<td>Center</td>
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<td>Inner radius, mm</td>
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<td>Composite Tube</td>
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<td>(US value $-$ actual value)/actual value</td>
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<td>Inner layer, 1% agarose</td>
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<td>Outer layer, 3% agarose</td>
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<td>Inner radius, mm</td>
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<td>Wall thickness, mm</td>
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Values are percentages. US, ultrasound.
The border between the bag lumen and esophageal wall was clear during distensions. The lining of the bag and mucosa could not be distinguished from each other in the ultrasound images. However, esophageal muscle wall could still be seen as a hypoechoic band, so the outer border of the esophagus could be outlined successfully even during deflation. The relationships between intrabag pressure, bag volume, and esophageal luminal CSA are shown in Fig. 3. Both relationships exhibit a linear tendency.

Figure 4 shows that a linear relationship also exists between the esophageal luminal CSA and bag pressure during isovolumic and isobaric distensions. The esophageal compliance in the circumferential direction, determined from the slope of the CSA-pressure relationship, tended to be higher for isovolumic distensions compared with isobaric distensions ($P = 0.066; 8.18 \pm 1.17$ and $4.97 \pm 0.97$ mm$^2$/mmHg in isovolumic and isobaric studies, respectively). It should be noted that during isovolumic distensions of 17.5 and 20 ml, the generated pressure exceeded 60 mmHg.

Effect of atropine. After administration of atropine, swallow-induced contraction amplitude was decreased by 72%, as determined from the pressure at 2 cm above the bag ($82.9 \pm 7.5$ vs. $22.0 \pm 2.2$ mmHg). Also, $L_0$ increased significantly, from $43.5 \pm 1.5$ to $49.8 \pm 2.3$ mm ($P = 0.021$), after atropine injection.

Circumferential stress-strain relationship during distension. In the isovolumic study, there was a linear increase in the strain, which ranged from $0.26 \pm 0.09$ to $1.77 \pm 0.24$ and a linear increase in the stress, which ranged from $7.54 \pm 0.70$ to $15.28 \pm 1.97$ kPa (Fig. 5A). In the isobaric study, there are linear increases in the strain and the stress, which range from $0.17 \pm 0.05$ to $0.95 \pm 0.12$ and from $2.73 \pm 0.48$ to $12.58 \pm 1.45$ kPa, respectively (Fig. 5B).

The stress-strain relationship shows a nonlinear trend that can be fitted by a polynomial (Fig. 6). The relationship is fairly linear if the zero point is neglected. The slope of the curve was 4.9 kPa ($r^2 = 0.94$) for the isovolumic study and 13.6 kPa ($r^2 = 0.95$) for the isobaric study.

Dynamic changes of stress and strain. The long period of esophageal distension with 15 ml volume or 40 mmHg pressure reveals that there are dynamic changes in the intrabag pressure, wall thickness, and esophageal luminal CSA during distension. We observed two characteristic patterns. Pattern 1 is characterized by a phasic increase in the bag pressure with an increase in the esophageal luminal CSA without any significant change in the wall thickness. During these instances, the wall stress and strain increase with the increase in the pressure and CSA (Fig. 7A). Pattern 2 consists of phasic changes in intrabag pressure with a decrease in the CSA and an increase in the wall thickness, suggesting contraction at the level of ultrasound probe. The esophageal strain decreases with the reduction in the CSA. The wall stress, however, either decreases or increases, depending on the degree of the increase in wall thickness. If the increase in wall thickness is small, the wall stress tends to increase while the strain decreases (pattern 2a). On the other hand, if the increase in wall thickness is large, the wall stress changes in parallel with the change in the strain, as shown in Fig. 7B (pattern 2b).

The distribution of the different patterns in nine subjects with sustained distension of 15 ml is as follows: pattern 1, three subjects; pattern 2a, two subjects; pattern 2b, one subject; mixture of patterns 1 and 2a, two subjects (converting from 2a to 1). In one subject, no pattern could be identified. The distribution of the different patterns in six subjects at 40 mmHg isobaric sustained distension is as follows: pattern 1, no subjects; pattern 2a, no subjects; pattern 2b, five subjects; no pattern, one subject.
In vitro validation. The in vitro studies confirm the accuracy of the measurements made by the ultrasound technique. The position of the ultrasound probe inside the bag and the angle between the probe and esophageal wall are important variables to ensure adequate ultrasound images. The design of our system for in vivo studies is such that it allows the placement of the probe in the center of the bag. Because the esophagus is a relatively straight tube, the angle between the transducer and the esophageal wall is unlikely to be large. The accuracy of the measurement is not compromised if the probe is slightly angled (<15°). This is especially true because the measurements of the wall thickness are made at four different positions around the circumference and then a mean thickness is obtained. We often observed that the esophageal wall was compressed by the aorta in one direction, resulting in an asymmetrical wall thickness along the circumference.

In vivo studies. The in vivo human studies show that the validated technique of manometry and bag ultrasonography allows measurement of pressure, visualization of the CSA of the esophagus, and the measurement of wall thickness. Our data show a linear relationship between esophageal pressure, CSA, wall stress, strain, and elastic modulus (parameter of wall rigidity) during esophageal distension. The magnitude of stress, strain, and elastic modulus vary depending on the degree and method of distension (isovolumic or isobaric). The reason for the difference between isovolumic and isobaric distensions is most likely related to the rate of the distension. Isovolumic distensions were performed by manual injection with an approximate speed of injection of 2 ml/s. On the other hand, isobaric distensions lasted until the CSA reached the maximum size, which usually required 20–30 s. Rapid and slow distensions caused different sensory and motor responses in human rectum (22, 25, 27). Sun et al. (27) reported that increasing the rate of inflation produced a graded increase in pressure at each volume and suggested that time-dependent relaxation of the smooth muscle is overcome at higher flow rates. Plourde et al. (22) also reported that pressure-volume curves for slow and rapid distensions are significantly different. It appears that different mechanoreceptors are activated during rapid and slow rectal distensions (17, 25). Ours is the first study in the esophagus that shows the dependence of stress and strain on the method of distension.

Fig. 5. A: relationship between Kirchhoff’s stress (σ) and bag volume (V) (∙) and between Green’s strain (ε) and bag volume (∙) in the isovolumic distension study (n = 9). The data are fitted by a least-squares fit: σ = 0.50V + 6.4 (r² = 0.93) and ε = 0.10V – 0.25 (r² = 0.99). B: relationship between esophageal wall stress and intrabag pressure (∙) and between strain and intrabag pressure (∙) in the isobaric distension study (n = 8). The data are fitted by a least-squares fit: σ = 0.21P + 1.01 (r² = 0.96) and ε = 0.016V + 0.005 (r² = 0.99).

DISCUSSION

In vitro validation. The in vitro studies confirm the accuracy of the measurements made by the ultrasound technique. The position of the ultrasound probe inside the bag and the angle between the probe and esophageal wall are important variables to ensure adequate ultrasound images. The design of our system for in vivo studies is such that it allows the placement of the probe in the center of the bag. Because the esophagus is a relatively straight tube, the angle between the transducer and the esophageal wall is unlikely to be large. The accuracy of the measurement is not compromised if the probe is slightly angled (<15°). This is especially true because the measurements of the wall thickness are made at four different positions around the circumference and then a mean thickness is obtained. We often observed that the esophageal wall was compressed by the aorta in one direction, resulting in an asymmetrical wall thickness along the circumference.

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Fig. 6. Relationship between Kirchhoff’s stress and Green’s strain for the isovolumic (∙) and isobaric (∙) studies. The data are fitted to a polynomial equation in the isovolumic and isobaric distensions; σ = 7.23ε³ – 23.58ε² + 27.56ε + 0.46 (r² = 0.99) and σ = –19.02ε³ + 20.08ε² + 11.35ε + 0.11 (r² = 0.97), respectively.
Phasic changes during distension. One of the strengths of our technique is that it provides measurements of the pressure, CSA, and wall thickness on a continuous-time basis. The data shown in Figs. 3–6 correspond to the point of maximum relaxation, following distension when the pressure is stable. During sustained esophageal distension, however, there are dynamic changes in the intrabag pressure, CSA, and wall thickness, which result in dynamic changes in the wall stress and strain. Distension-induced contraction at the site of bag distension is consistent with Orvar’s study in humans (18) that measured the luminal CSA and the intraballoon pressure. Paterson et al. (20) also reported phasic changes in the intraballoon pressure during distension in the opossum esophagus; they described that the secondary contraction above the balloon squeezes the balloon and results in the fluctuation of the intraballoon pressure. In our study, during some of the long distensions, these pressure changes were associated with changes in the luminal CSA and the wall thickness of the esophagus and others were not. The changes in wall thickness and luminal CSA as a result of contraction around the balloon have never been described before. These changes in the wall thickness resulted in different patterns of stress and strain, although the distending stimulus was the same. The dynamic nature of the stress-strain relationship in response to distension has not been described before. The significance of such dynamic relationships in normal healthy subjects is not known but must be important to maintain the homeostasis of the esophageal wall during various physiological functions. According to our classification, pattern 1 shows no contraction but pattern 2 shows active contraction. Furthermore, within pattern 2, pattern 2b reveals stronger contraction than pattern 2a. A notable point is that the esophagus can contract around a bag in the presence or absence of contraction above the bag (pressure change at 2, 7, and 12 cm above the bag were monitored in our study). This contraction appears to be phasic. In future studies, analytical methods should be used to quantitatively characterize these waveforms during distension.

Comparison with the other works. Biancani et al. (2, 4) studied the biomechanics of the LES and esophagus in animals and humans. They measured pressure by using manometry, and the wall thickness was measured after killing the animals or at the time of autopsy. One of the assumptions made by this group was that there is no difference in the wall thickness in vitro compared with the in vivo state. However, ultrasound recordings of the esophagus show that the wall thickness in vivo strongly depends on the tone of the muscle (9, 14, 21). Furthermore, during LES and esophageal contraction there is an increase in the thickness of the esophageal wall. Mayrand and Diamant (12) studied the compliance of the esophagus by using a barostat technique. Although compliance is an important parameter of the wall biomechanics, it does not provide a measure of stress and elastic modulus (wall rigidity), which are two important variables that may directly relate to motor and sensory functions of the esophagus. The impedance planimetry, used by Gregersen and his colleagues (5), provides information on pressure and CSA of the esophagus but does not measure the wall thickness. The combined high-frequency A-mode ultrasound-impedance planimetry designed by Jorgensen et al. (8) addressed this limitation and offers the measurement of wall thickness in vivo. Jorgensen et al. formulated a uniaxial constitutive relationship in terms of Cauchy’s definition of stress and strain in the strain range of ~0–200%. Cauchy’s strain, however, should only be used in small deformation (strain < 10%). Hence, their use of the small strain measure is inappropriate. In the present study, we used Green’s strain and Kirchhoff’s stress, which are the appropriate and complementary pair for large deformation. Hence, we have further improved Jorgensen’s methodology by the use of B-mode ultrasonography and improved the theoretical formulation to yield novel data on the phasic deformation of the human esophagus during distension.

Model assumptions. A number of assumptions were invoked in the computations of mean stress and strain in the wall of the esophagus that deserve mention. The hoop stress computed in Eq. 2 is defined as the force/area in the circumferential direction. The underlying assumption is that there exists an equilibrium of forces, i.e., the esophagus is either in a static condition or, if it is in motion, the motion occurs with constant velocity (the inertial forces due to acceleration are considered negligible). It is obvious that the esophagus undergoes considerable geometric changes during contraction. The most reliable computations of stress are either at the point of complete relaxation or contraction at which the pressure is most stable. The values of stress shown in Fig. 6 were made at the point of minimum contraction at which the assumption of static equilibrium, i.e., no motion, is justified. It should be noted that the wall stress given by Eq. 2 consists of the passive (due to the elastic and viscoelastic properties of the tissue) and active (due to the muscle contraction) stress. One of our future goals is to separate the passive and active components.

Mechanical strain describes the deformation of a material. Any measure of deformation or strain must be made in reference to the zero-stress state of the...
tissue (3). It is impossible to determine the zero-stress state in vivo because even if the muscles are fully relaxed, the tissue bears residual strain and stress (7). To reveal the zero-stress state of the esophagus, a sufficient number of cuts must be made to remove all of the internal stresses and strains. Because this is not possible in human in vivo experiments, an estimation of the zero-stress state has to be made. Some investigators have used small distension at small pressure as the zero reference (23, 26). Recently, Patel et al. (19) calculated the esophageal strain based on zero pressure in vivo. This unloaded (zero pressure) state of the esophagus, however, still contains stress and strain due to muscle tone. Mayrand et al. (12) show, in human studies using the barostat technique, that smooth muscle tone has an active component that can be reduced by a smooth muscle relaxant such as amyl nitrate. In the present study, we used atropine to reduce the majority of tone. Atropine increased resting esophageal circumference by ~15% and decreased the amplitude of swallow-induced esophageal contraction by ~70% in the smooth muscle esophagus. This implies a decrease in tone and diminished contractility, suggesting that there is tonic cholinergic input to the esophageal body. The use of atropine more closely approximates the zero state. Even in the complete absence of tone, however, some residual strain still exists in the esophagus. Gergersen et al. (7) have previously shown that the residual strains in the guinea pig are approximately -0.3 and 0.1 at the inner and outer surfaces of the esophagus, respectively. Hence, the submucosa and muscle experience residual compression and tension, respectively. The calculation of strain in the present study must take into account the residual strain in the human esophagus when those data become available.

Elasticity parameters. A number of elasticity parameters have been previously defined to characterize the mechanical properties of the esophagus. Compliance is defined as the change in luminal dimension divided by the corresponding change in pressure. This parameter merely expresses the differences in luminal dimensions between pressure steps. Hence, it does not take into account the actual degree of stretch or the wall thickness. Pressure elastic modulus, a parameter for the wall stiffness, has been calculated using the equation \( r \times dP/dr \), where \( r \) is the radius and \( dP/dr \) is the changes in balloon pressure and radius between two consecutive steps. Clearly, this is more advantageous than compliance because it considers the degree of stretch. However, its limitation is that it does not account for the changes in wall thickness. The slope of the stress-strain relationship (tangent modulus) takes into account the degree of stretch and the wall thickness (3). Hence, it is a preferable measure of wall rigidity. For a material with a linear stress-strain relationship (i.e., Hookean material), the tangent modulus is called Young’s modulus. Our results show a linear stress-strain relationship in the range of pressure (0–60 mmHg) and volume (5–20 ml) of the distensions used. The relationship may become nonlinear in the low and high regimes of stress and strain.

Critique of methods. Our bag accommodates a maximum volume of 25 ml before the bag itself contributed to the intrabag pressure. We found that with this size bag, during volume distension of <5 ml, the bag was somewhat folded and prevented complete visualization of the luminal CSA. For this reason, we chose the outer esophageal circumference for calculations of strain rather than inner esophageal circumference or mid-esophageal circumference, which would theoretically be more appropriate.

During constant-volume distension, the manual injections were completed in shorter time than during isobaric distension. The former may result in a stronger secondary peristaltic contraction above the bag than in the constant-pressure distension study. We computed a time for calculation of the esophageal stress and strain when the pressure was at the lowest value. There remains the possibility, however, that this point did not reach a true minimum pressure.

Significance of study. How does measurement of stress, strain, and wall rigidity help in our understanding of the esophageal function? Mechanical properties of the esophagus are an important determinant of tone, peristaltic reflexes, bolus transport, mechanoreceptor responses, and sensory perception (6). The stress-strain relationship is an expression of the mechanical properties of the tissue. The roles of stress and strain in the cardiovascular system have a firm foundation. It is well known that stress and strain regulates the growth and remodeling of blood vessels. For example, chronic exposure of the vessel wall to tensile stress and strain (hypertension) leads to growth and “vascular remodeling.” In large-conduit arteries of the systemic circulation, increased pressure and tensile stress is associated with smooth muscle hypertrophy, increased media deposition of collagen, and destruction of elastin fibers that leads to arterial stiffness, decreased compliance, and increased flow impedance (11, 16). According to Laplace’s equation, the mean circumferential stress in the vessel wall is directly proportional to the blood pressure and inversely proportional to the vessel radius and thickness ratio. It appears that the vessel wall thickness-to-radius ratio increases in proportion to the increase in blood pressure so that the stress remains constant. We recently found that, during esophageal contraction, as the esophageal pressure increases so does the muscle thickness, which would indicate that the wall stress remains relatively constant during contraction (21). Furthermore, the remodeling of the esophagus in various diseases (e.g., increase in the thickness of muscularis propria in nutcracker esophagus and diffuse esophageal spasm (10, 15)) may well be in accordance with the “uniform stress” hypothesis. A homeostatic state of stress must exist in an organ that is closely regulated, and when homeostasis can no longer be maintained the organ may fail. Congestive heart failure, aneurysm, and achalasia of the esophagus are examples of the failure of the heart, blood vessels, and esophagus, respectively. The mechanical characterization of the esophagus in the normal subjects is thus crucial to the understanding of the normal
function of the esophagus and will serve as a reference for understanding various pathological states. Furthermore, distension of the esophagus and other viscera is a known stimulus that induces esophageal and visceral sensation (13, 24). The relationship between stress, strain, and esophageal sensation is not known.

REFERENCES


