Static and dynamic operating characteristics of a pericardial balloon

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Hamilton, Douglas R., Gwyneth Devries, and John V. Tyberg. Static and dynamic operating characteristics of a pericardial balloon. J Appl Physiol 90: 1481–1488, 2001.—Previously, we developed a balloon transducer to measure the constraint of the pericardium (i.e., pericardial pressure) on the surface of the heart. It was validated physiologically in that it was shown to measure a pressure equal to the difference between the left ventricular end-diastolic pressure measured before and after pericardiectomy at the same left ventricular volume. To define its static operating characteristics, we loaded the balloon nonuniformly with weights that covered fractions of the balloon surface and found that the balloon accurately recorded the average stress if the stress was applied over at least 23% of its surface. To test its performance when curved, we placed it in large and small cylinders (minimum diameter 31 mm) and found that the balloon accurately recorded the stress. To define its dynamic operating characteristics, we applied sinusoidal stresses and found that its frequency response was limited only by that of the connecting catheter. When better dynamic response is required, we introduce a micromanometer-tipped catheter to obtain a unity-gain frequency response that is flat to 200 Hz.

pericardium; pericardial pressure; transducer; frequency response; epicardial stress

A CENTURY AGO, Barnard (5) concluded that the pericardium is a significant constraint to filling of the heart. Many early investigators also observed that the parietal pericardium exerts a significant stress on the epicardium, but various transducers and catheters used to measure this constraint have yielded conflicting results (14, 17, 20, 24, 30).

Scalar terms such as “pressure” have been used to describe the stresses that the pericardium imposes on the epicardial surface. If one considers the tangential stress (shear stress) at the interface of these two structures to be minor, then the remaining radial stress might be considered to be scalar in nature, because the total vector stress on the epicardium will be normal to both surfaces, emulating liquid behavior over small elemental areas. The same situation can exist if a sufficient amount of liquid is present in the pericardial space to separate the epicardial and pericardial surfaces such that the distribution of radial stresses behaves hydrostatically (25).

Attempts have been made to measure pericardial pressure directly by using open-end catheters (4, 14, 20, 30), air-filled balloons (9, 17, 27), and liquid-filled balloons (12, 25). As measured by these devices, pericardial pressure has ranged from zero to nearly the cavitary pressure of the chamber being investigated, indicating the need for a standard in transducer design and calibration.

In 1985, Smiseth et al. (25) showed that pericardial pressure was strikingly dependent on the method of measurement used. They confirmed that the conventional open-ended catheters used to measure pericardial constraint in earlier studies registered a negligible pressure; however, they also showed that a balloon adapted from Holt et al. (12) recorded a hemodynamically significant pressure. (Since that time, the design and specifications of the balloon have not been changed.) The pressures measured from these flat balloons in surgical patients (7, 10, 29) were later shown to follow the blood volume-induced changes in right atrial pressure, in both direction and magnitude. As Tyberg et al. (28) originally proposed, the magnitude of pericardial constraint may be determined by comparing the (vertical) shift in the left ventricular (LV) end-diastolic pressure-volume relationships before and after pericardiectomy. This rationale for determining the true magnitude of pericardial constraint has been adopted by several other investigators (2, 3, 11).

The purpose of this investigation was to provide a systematic description of the fabrication technique and the static and dynamic operating characteristics of the balloon transducer used in several previous studies (6, 10, 15, 25, 29). In addition, we will describe a simple modification of the balloon that provides higher frequency response and fewer motion-induced artifacts.

METHODS

Balloon fabrication. The pericardial balloon is fabricated by using a 7 cm × 4 cm flat Silastic sheet (0.25 mm thick; compound AR131; Armet Industries, Concord, Ontario, Canada) wrapped around a 14-cm length of medical grade Silas-
tic tubing (1.58 mm ID; 3.18 mm OD; Dow Corning, Midland, MI, catalogue no. 602-285). The tubing has three 2.0 mm^3 1.5 mm holes, spaced 5.0 mm apart, cut into its side starting 7.5 mm from the end (see Fig. 1). The Silastic sheet is then wrapped around the tubing, and the opposing surfaces are bonded by using silicone adhesive (Dow Corning, catalog no. 891). The fabrication procedure is standardized by using a metal mold (schematic diagram available from the National Auxiliary Publications Service) that allows the balloon to be cured under slight pressure (10 mmHg). This method standardizes the internal volume and geometry. The opposing surfaces to be bonded are cured at 50°C for 24 h and then removed from the mold and left to stand at room temperature and out-gas for another 24 h. The edges of the balloon are trimmed to standard dimensions (30 mm \times 30 mm; a smaller, 14- \times 12-mm balloon has also been fabricated). The advantages of using Silastic sheets and silicone adhesive are the long shelf life after fabrication, nonreactivity to body tissues (8), flexibility, and resistance to fatigue and repetitive gas sterilizations. For most experiments, the balloon is attached to a length of 7-Fr angiographic catheter material, cut as short as convenient. Recently boiled room-temperature water is introduced into the balloon, and air bubbles are purged. Operating volumes range from 0.70 to 0.90 ml.

To minimize distortion due to catheter motion and to increase frequency response, we introduce a 3-Fr micromanometer-tipped catheter (model PR-249 3F Mikro-tip catheter, Millar Instruments, Houston, TX) through the connecting catheter into the balloon (the micromanometer-tipped catheter remains within the connecting catheter, thus preventing exposure to direct compression). Because the micromanometer-tipped catheter is susceptible to baseline drift, the connecting catheter is also attached to a conventional strain-gauge pressure transducer (model P23 Db, Statham-Gould, Oxnard, CA), which allows an accurate measurement of mean pressure. The mean pressure of the micromanometer-tipped catheter is corrected to be equal to that measured through the 7-Fr catheter.

**Static calibration: Uniform stress application.** For convenience and reproducibility, we designed a compact balloon calibrator to apply stresses normal to the balloon surface using a thin, unstressed, compliant Silastic membrane (Fig. 2). The thin membrane (the same material used in the pericardial balloon) was attached to a Plexiglas cylinder (12.7 cm OD; 11.4 cm ID) so that it was unstressed but did not present any fold or redundancy to the balloon or to its supporting end plate. The other end of the cylinder was sealed with a Plexiglas end plate (15.2 cm, diameter; 1.27 cm, thickness). By placing the Silastic membrane over the pericardial balloon, which lay on the flat end plate, the desired configuration was achieved. The end plate was held against the thin membrane by use of three brass screws. A small trough was machined into the end plate, permitting passage of the tubing for connection to a pressure transducer. The pressure inside the calibrator could be manipulated and measured through a side port. Because the Plexiglas was clear, it was possible to observe the balloon to ensure its proper position. The end plate had several small holes drilled through its surface to allow air to escape. This ensured that the only stress applied to the balloon was through contact with the Silastic membrane and the supporting end plate. To permit testing under negative pressure, an alternate end plate was fabricated. This end plate had only one hole, which allowed air to escape

![Fig. 1. Schematic diagrams of the pericardial balloon. A, connecting tubing; B, transducing contact area (x \times y); C, border (width = z) to prevent buckling; and D, side ports cut in tubing to communicate with balloon. For the balloon described in this study, x = 30, y = 30, and z = 5 mm. For atrial studies in large animals and for ventricular studies in cats and large-animal neonates, a smaller balloon (x = 12, y = 14, z = 5 mm) has also been fabricated.](http://jap.physiology.org/)

![Fig. 2. Exploded-parts view of the static balloon calibration device showing the Plexiglas pressure chamber (A), Silastic membrane (B), membrane support ring (C), and balloon support disk (D).](http://jap.physiology.org/)
during positive pressurization; the hole was then sealed and the pressure in the test chamber was lowered to expose the balloon to negative stresses. Details of the calibration procedure are given in Appendix A.

Static calibration: Nonuniform stress application. A benchtop experiment was performed to investigate whether the pericardial balloon would accurately record an average stress, even if the load had been applied nonuniformly. The effects of nonuniform loading were examined by placing rectangular blocks of different areas (1.0, 2.1, 4.0, 6.1, and 8.0 cm$^2$) at the center of the balloon's transducing contact surface (see Fig. 1) while the balloon was lying on a flat surface. These blocks were loaded with weights that were adjusted to apply identical stresses (i.e., force per area of the block). Arbitrarily, we chose stresses equal to 18 mmHg (23,940 dynes/cm$^2$) and 32 mmHg (42,560 dynes/cm$^2$). For example, to achieve a stress equivalent to 32 mmHg, the 2.1 cm$^2$ block was loaded with 91.1 (43.4 × 2.1) gm, the 4.0 cm$^2$ block was loaded with 173.6 (43.4 × 4.0) gm, and so on. The pressures measured by the balloon were plotted against the areas of the blocks when so loaded. If the measured pressure was proportional to the area of the balloon surface that was covered by the loaded block, it was concluded that the balloon had spatially integrated (i.e., averaged) the applied stress appropriately.

Static calibration: Stress applied over a curved surface. Bending a balloon might introduce Laplacian stresses on the membranes, thereby exaggerating the recorded pericardial pressure. The calibration was verified by placing the pericardial balloon within different Plexiglas cylinders (internal diameters 31, 38, and 50 mm) and inflating a thin-walled cylindrical balloon (i.e., a condom) to compress it against the inside wall of the cylinder (see Fig. 3). The surfaces were flushed with water, and the condom was inflated to different measured pressures. According to the pleural-pressure measurement technique developed by Lai-Fook et al. (16, 31), the stress that the condom exerted on the inner surface of the cylinder was measured by use of a pressure transducer (Model P23 Db, Statham-Gould) connected by a liquid-filled tubing to a hole in the wall of the cylinder at the same hydrostatic level as the pericardial balloon. The pressure measured by the pericardial balloon was compared with the condom pressure and that measured by the wall transducer. Later, the unconstrained condom was inflated and the distending pressure was plotted as a function of condom diameter (measured using vernier calipers), to allow estimation of the transmural pressures required to achieve condom diameters equal to the inside diameters of the cylinders.

Dynamic calibration. To determine the balloon’s frequency response, we constructed a dynamic calibration chamber using a high-fidelity speaker (model SW6025G Alpha Subwoofer, Heco-Hennel and, Germany). The speaker was driven by two 100-W, direct-current coupled amplifiers arranged in a class-B push-pull configuration (model SM-100, OPAMP Labs, Los Angeles, CA), which amplified a sinusoidal input signal (model 180 LF sweep generator, Wavetek, San Diego, CA). The static calibrator was modified to accommodate the speaker, which imposed a time-varying pressure onto the membrane overlaying the pericardial balloon (schematic diagram available from the National Auxiliary Publications Service). The oscillating chamber pressure was measured by using a micromanometer-tipped catheter (model PR-279 8F Mikro-tip, Millar Instruments) introduced through a side port. The signal from the sweep generator was compared with the pressure measured in the chamber by using specially designed circuitry to create an error signal that was used for negative feedback to the power amplifier of the speaker. This sinusoidal pressure-generating system kept total harmonic distortion of the chamber-pressure waveform to <5% over frequencies ranging from 2 to 200 Hz (peak sinusoidal pressures up to 20 mmHg). The calibration chamber was designed to be pressurized to 60 mmHg (i.e., the mean chamber pressure could be 40 mmHg on which a ±20 mmHg sinusoidal pressure could be modulated). To measure the stresses effectively being imposed on the balloon by the membrane, a load cell (diaphragm diameter 25 mm; model P23-BB venous pressure transducer, Statham, Hato Rey, Puerto Rico) was placed flush with the surface of the balloon-support end plate, immediately beneath the position to be occupied by the balloon. The load-cell output was also measured with the balloon in place to assess the dynamic effect of the presence of the balloon. If the relation of chamber pressure to load-cell “pressure” (i.e., stress) was unaffected by the presence of the balloon, it was concluded that the balloon did not alter the dynamic characteristics of the system and balloon pressure would track chamber and load-cell pressures. The pressure recorded from the balloon and that from the load cell were compared with the chamber pressure in terms of amplitude and phase lag.

RESULTS

Static calibration: Uniform stress application. More than 90% of the balloons fabricated using the mold were found to pass the static calibration procedures. A typical calibration plot is shown in Fig. 4, which illustrates the linearity of these devices. The difference between the calibrator and balloon pressure never ex-
ceeded 1 mmHg over the range of pressures. When properly calibrated, 10 mold-fabricated balloons were found to have volumes of $0.775 \pm 0.054$ ml (means $\pm$ SD) at which they were unstressed.

Static calibration: Nonuniform stress application. Figure 5 shows the results of the nonuniform loading test. When rectangular blocks of different sizes were loaded with weights such that stresses equivalent to 32 or 18 mmHg were applied, the pressures recorded from the balloon were proportional to the areas of the blocks. When a block smaller than 23% of the transducing contact surface (e.g., 11%) was used, the balloon underestimated the applied stress. However, the balloon transducer accurately recorded the average stress when that stress was applied over at least 23% (2.1/9.0) of its transducing contact surface.

Static calibration: Stress applied over a curved surface. Figure 6 shows that, when the pericardial balloon was placed within cylinders of various sizes, the recorded pressure increased 1:1 with the inflation pressure of the condom. The X-axis offset can be approximately accounted for by the fact that the unconstrained condom required pressure to attain the diameters of the cylinders: 1, 5, and 10 mmHg (note x-intercepts) to achieve diameters of 31, 38, and 50 mm. We concluded that, if the balloon is placed on a 1-dimensionally curved surface with a radius of curvature as small as 15.5 mm, it still measures contact stress accurately.

Dynamic calibration. In Fig. 7 (top), the amplitude ratio (pericardial balloon pressure/chamber pressure) is plotted against the frequency of the sinusoidal input function. A typical second-order, underdamped response with a peak resonance and a 180° phase shift at $\sim 25$ Hz is seen (the falloff was linear when amplitude ratio and frequency were plotted on logarithmic scales). Because there was always a 1:1 relationship between the chamber pressure and the load-cell “pressure” (data not shown), it was clear that the calibrator membrane did not modify the time-varying chamber pressure. The frequency response of the high-fidelity balloon was flat to 200 Hz (only data to 100 Hz shown), we concluded that the limitation in the frequency response of the pericardial balloon was determined by the catheter. As shown in Fig. 7, bottom, the phase response for the pericardial balloon shifted from...
there was no phase shift when using the high-fidelity balloon (data not shown).

Over this frequency range, the cardiac balloon shifted from 0 to 180° with a phase shift of 90°.

DISCUSSION

The simple tests reported in this paper demonstrate that the pressure recorded from a flat, liquid-filled, pericardial balloon transducer accurately reflects the stress applied to its surface. This is true not only when a static stress is applied uniformly to the balloon but also when the stress is applied over as little as 23% of its area. It is also true when the balloon is deformed by placing it on a surface more curved than the free wall of the canine LV. Although the frequency response of the standard balloon transducer is limited by that of the connecting catheter, the standard balloon can be modified by inserting a micromanometer-tipped catheter inside the balloon, extending its frequency response to 200 Hz, and reducing its susceptibility to catheter motion-induced artifacts. Therefore, in addition to the fact that the balloon transducer records a pressure equal to the constant-volume shift in LV end-diastolic pressure (28), these results indicate that there is no reason to doubt that the balloon transducer can be used to measure epicardial radial stress (i.e., "pericardial pressure") reliably throughout the cardiac cycle.

Although we and others have filled pericardial balloons with water or saline (7, 12), other pericardial investigators have filled their balloons with air (3, 11, 24), following the precedent of those who used balloons to measure pleural or esophageal pressure (13, 19). Any fluid (air, water, oil, etc.) can be used, but, because of compressibility, air-filled balloons perform somewhat less well than water-filled balloons in terms of nonuniform loading and, because of viscosity, oil-filled balloons perform less well in terms of frequency response (data not shown). However, as discussed in detail in Appendix B, liquids with bloodlike specific gravities are very advantageous because they allow accurate assessments of transmural pressure, regardless of the position of the balloon.

Conventional static calibration techniques (18) load the balloon with a uniform stress distribution. We developed a more rigorous test in which loading was applied nonuniformly, probably much less uniformly than could occur within the pericardium. We applied equal, known stresses over varying fractions of the transducing surface of the balloon. We reasoned that, if a given stress were applied over half the surface, for example, the proper response would be one-half the pressure that would be registered if that same stress had been applied uniformly over the entire surface. We found that the balloon performed this spatial integration very well, probably because of its construction. The loaded rectangular block would have caused Laplacian stresses to develop in both surfaces. If those stresses had not been opposed, a pressure less than the predicted pressure would have been recorded from the balloon. Because the measured pressure was equal to the predicted pressure, those Laplacian stresses must have been opposed by the rigidity of the borders of the balloon, which caused pressure to increase to the predicted level. The borders of a pericardial balloon must be rigid enough so they resist the buckling that would tend to occur because of the tension in the Silastic membranes created by nonuniform loads, yet be thin enough to be flexible (1, 18).

It has been suggested that, because of the curvature of the LV surface, the balloon must record an artifici-
ally high pressure, even though it had been calibrated to be accurate when flat (21). The surface of the LV is hemiellipsoidal and, therefore, curved in two directions. We made several attempts to create a test chamber with an elliptical surface. However, we abandoned the effort because we were unable to specify the applied stress unequivocally, because of the complexity of the stresses. We then chose an alternative strategy: to demonstrate that the balloon performed accurately in a cylindrical calibrator with a relatively extreme (albeit one-dimensional) curvature in which the imposed stresses could be defined (16). We observed that

![Fig. 7. Dynamic calibrations of the regular and high-fidelity pericardial balloons. Top: amplitude ratio (pericardial balloon pressure/chamber pressure) is plotted against frequency of the sinusoidal input function. For the regular balloon (larger circles), a typical second-order, underdamped response with a peak resonance at ~25 Hz is seen (the falloff is linear when plotted log-log). For the high-fidelity balloon (smaller circles), the amplitude ratio is ~1 at all frequencies <100 Hz. Bottom: phase response for the regular pericardial balloon shifted from 0 to 180° with a phase shift of 90° corresponding to the peak resonance. Over this frequency range, there was no phase shift when using the high-fidelity balloon (data not shown).](https://jap.physiology.org/Downloadedfrom)
the balloon was accurate even in the cylinder with the smallest radius of curvature; 15.5 mm is less than either the short- or long-axis radius of curvature of the LV free wall of a 20-kg dog and much less than those dimensions in the adult human. Thus we conclude that there is no significant error due to curvature when the balloon transducer is used to measure pericardial pressure over the LV free wall.

Predictably, we found that the frequency response of the standard pericardial balloon was not better than that of the connecting 7-Fr catheter. Figure 7 indicates that the 20-Hz component of the pressure waveform was approximately doubled in magnitude. We also evaluated the output of the high-fidelity balloon to assess the frequency response required to measure the fastest events in the cardiac cycle (i.e., the early systolic peak in pericardial pressure accurately, the high-fidelity balloon is required (unfiltered pressure, 4.56; 40-Hz filter, 4.52; 30-Hz filter, 4.34; 20-Hz filter, 4.12; and 10-Hz filter, 3.77 mmHg).

In conclusion, we have performed static and dynamic calibrations of a Silastic, liquid-filled, pericardial balloon transducer. Static performance was satisfactory, even when loading was markedly nonuniform or when the balloon was curved. Dynamic performance was limited only by the transmission characteristics of the

filtering and, thus, we conclude that the frequency response of the standard pericardial balloon transducer is sufficient to measure end-diastolic transmural pressure accurately. However, when faster events are to be studied, the high-fidelity adaptation of the transducer is satisfactory and convenient.

In addition to allowing the study of faster events, the insertion of a micromanometer-tipped catheter into the balloon minimizes motion-induced artifacts (i.e., “catheter-whip”). Although the connecting catheter might still generate such artifacts, the micromanometer in the balloon will not detect them because the mechanical impedance of the balloon is very low. We were able to demonstrate that “artifacts” generated in the catheter by tapping, which were conspicuous when measured using the external transducer, were undetectable using the micromanometer-tipped catheter. The problem of micromanometer baseline instability can be overcome by recording pressure through the lumen and adjusting the mean values to be equal either manually, before the data are recorded, or retrospectively, during analysis, as we did.

In conclusion, we have performed static and dynamic calibrations of a Silastic, liquid-filled, pericardial balloon transducer. Static performance was satisfactory, even when loading was markedly nonuniform or when the balloon was curved. Dynamic performance was limited only by the transmission characteristics of the...
connecting catheter. When higher frequency response is required or motion-induced artifacts are to be minimized, a convenient high-fidelity modification is available.

APPENDIX A: DETERMINATION OF BALLOON VOLUME

Recently boiled room-temperature water is introduced into the balloon, and all air bubbles are purged. The balloon is positioned in the calibration chamber, and chamber pressure is increased to 5 mmHg to expel the free air from the space between the membrane and the end plate. The calibration chamber pressure is then returned to atmospheric pressure. Zero balloon volume is established when liquid is withdrawn until the balloon surfaces form a dimple over the side ports of the internal tubing (suction of approximately −5 mmHg is required for this to occur). The chamber pressure is increased to ~30 mmHg, and the volume of liquid introduced into the balloon is adjusted until the balloon pressure equals the chamber pressure within ±0.5 mmHg. The balloon volume is fixed at the value determined by the previous step, and balloon error is measured over chamber pressures ranging from −40 to +50 mmHg. If the balloon error is found to be >1 mmHg, the balloon is considered flawed and is discarded.

APPENDIX B: COMPARISON OF AIR- AND LIQUID-FILLED BALLOONS

Assuming that blood has a specific gravity of 1 gm/cm³, that the hydrostatic equivalent of LV diameter is equal to 4 mmHg (5.44 cmH₂O = 4 mmHg) and that the true transmural LV end-diastolic pressure is 2 mmHg at every point on its circumference, Fig. 9 shows that air- and liquid-filled balloons yield different values of transmural pressure, depending on the position of the balloon. If absolute LV intracavitary pressure is assumed to be 10 mmHg at the mid-LV reference plane, the absolute pressure must be 8 mmHg at the top of the cavity and 12 mmHg at the bottom, because of the weight of the blood. An air-filled balloon at the top of the LV (A in Fig. 9) would record 6 (8 − 2) mmHg, at the midplane (B) 8 (10 − 2) mmHg, and at the bottom (C) 10 (12 − 2) mmHg. On the other hand, a liquid-filled balloon would record 8 mmHg at every location. This is because the siphon effect of the column of water contained in the connecting catheter automatically adds or subtracts an appropriate hydrostatic correction after the transducer has been zeroed at the mid-LV reference plane. Because it is difficult to ascertain the position of the balloon precisely and at all times, it is difficult to correct the measurements from an air-filled balloon and, thus, to achieve accurate estimations of transmural pressure at any time in the cardiac cycle.

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