**Effect of Vessel Compliance on the In-Vitro Performance of a Pulsating Respiratory Support Catheter**

Intravenous caval respiratory support (or membrane oxygenation) is a potential therapy for patients with acute respiratory insufficiency. A respiratory support catheter is being developed that consists of a bundle of hollow fiber membranes with a centrally positioned pulsating balloon to enhance gas exchange. This study examined the influence of vessel compliance on the gas exchange performance of the pulsating respiratory support catheter. Polyurethane elastic tubes were fabricated with compliance comparable to that measured in bovine vena cava specimens. The gas exchange performance of the respiratory catheter was studied in an in-vitro flow loop using either the model compliant tube or a rigid tube as a “mock” vena cava. Balloon pulsation enhanced gas exchange comparably in both rigid and model compliant vessels up to 120 bpm pulsation frequency. Above 120 bpm gas exchange increased with further pulsation in the rigid tube, but no additional increase in gas exchange was seen in the compliant tube. The differences above 120 bpm may reflect differences in the compliance of the elastic tube versus the natural vena cava.

**Keywords:** Intravascular Oxygenator, Artificial Lungs

**Introduction**

Intravenous respiratory support is a potential alternative to mechanical ventilation under development for patients with acute respiratory insufficiency [1–3]. The process involves placing a bundle of hollow fiber membranes, as used in standard extracorporeal blood oxygenators, into the vena cava through insertion at the common femoral vein. The fiber membranes are manifolded and connected to an external oxygen source that provides for continual oxygen “sweep” gas flow through the fiber lumens. Oxygen diffuses out of the fibers into the venous blood, while CO2 diffuses out of venous blood into the fibers, thus providing supplemental respiratory support independent of the lungs. The extrapulmonary gas exchange reduces the need for mechanical ventilation, and thus can also reduce the associated progressive lung damage that can further exacerbate respiratory insufficiency in the patient [4]. The respiratory support catheter being developed in our laboratory uses a centrally positioned balloon within the fiber bundle [5,6,7]. Rapid balloon pulsation promotes blood flow past the fiber surfaces, reduces diffusional boundary layer thickness, and increases the gas exchange rate of the respiratory catheter. In recent ex-vivo tests of the catheter [7], balloon pulsation increased gas exchange for both O2 and CO2 by 50-500 percent depending on balloon pulsation rate and blood flow rate past the catheter.

Rational design development of a respiratory support catheter requires ongoing device characterization and testing on the bench, usually within a “mock” vena cava flow loop, in which the device is placed within a cylindrical vessel (tube) and perfused at physiological flow rates [6]. As design changes are made in fibers and fiber bundle configuration, pneumatic pathways, and balloon geometry for our catheter, the mock vena cava test allows assessment of these changes under physiologically relevant conditions of controllable known flow rates and defined fixed vessel geometry. Most often rigid tubing has been used as the “mock” vena cava in a variety of in-vitro bench studies of other intravascular oxygenators [8–10]. Similar to these other studies the “mock” vena cava within which the catheter lies in our bench test is Tygon tubing, which is chosen to reflect the average size of the adult human vena cava, but unlike the vena cava is essentially non-compliant (rigid) in the pressure range involved. Clearly, compliance of the host vessel per se may affect the gas exchange characteristics of a respiratory support catheter with a pulsating balloon. Balloon pulsation and the associated pressures swings can produce wall motion and diameter oscillations in a compliant vessel. These oscillations may impact on the flow patterns surrounding the respiratory catheter and as a result may affect gas exchange achieved by the catheter.

The purpose of this study was to assess in an in-vitro test whether physiologically relevant compliance of the host vessel markedly influences the gas exchange performance of a pulsating respiratory support catheter. Custom fabricated polyurethane tubing was used as the mock vena cava in the bench test system to tailor the compliance of the host vessel to that of the vena cava. Because the literature appears scant for direct measurements of normal vena cava compliance, or compliance of other large veins, we began this study with measurements of vessel compliance in bovine vena cava specimens.

**Materials and Methods**

The overall methods involved measuring the compliance of bovine vena cava specimens, fabricating an elastic tube with comparable compliance, and measuring the gas exchange performance of the respiratory support catheter in the model compliant tube along with the standard rigid tube as mock vena cava.

**Compliance Measurement.** Unbranched venous segments of approximately 6 cm were obtained from five calves at necropsy by dissection of the inferior vena cava immediately below the right atrium. The external diameter of the vessel segment was measured with a caliper before removal from the animal. Prior to dissection a fixed clamp was attached to the vessel and the vessel was...
marked with two reference ink dots a measured distant apart so that vessels could later be tested at their in-vivo length.

Vessel compliance was measured by placing the dissected vessel segment within the static holder of the setup shown in Fig. 1. Each end of the vessel was tied onto a threaded connector and the connectors were attached to two hosepipe clamps of the static holder, attached to one another by two long screws. The screws were adjusted until the vessel was at its in-vivo length as measured by the reference ink dots. A plug was placed at one connector and the other connector was attached to a piece of tubing. The static holder was then submerged in a solution of isotonic saline, heparin $\sim 10 \text{ ml/1000 ml}$, and sodium bicarbonate $\sim 8 \text{ ml/1000 ml}$ saline at $37^\circ \text{C}$, and pH 7.4. The tubing was then attached to a fitting in the chamber that enables the vessel to be filled and emptied.

Before determining the pressure-volume relationship of the vena cava specimen, the vessel was filled with a volume of fluid comparable to its in-vivo volume ($V_0$) as estimated by the measured in-vivo diameter and length; this is also approximately the volume at which the transmural pressure just rises above zero (i.e., the resting volume). The pressure-volume relationship was then determined by infusing buffer in $\Delta V = 0.5 \text{ ml}$ increments until the transmural pressure reached 40 mmHg (5.33 kPa), at which time the vessel was returned to $V_0$ and buffer was removed in 0.5 ml increments until the transmural pressure reached $-5 \text{ mmHg}$ ($-0.67 \text{ kPa}$). Transmural pressure was measured using a Validyne differential pressure sensor (CD 379, Northridge, CA) with one port in the vessel lumen and the other in the bath at the same elevation. The pressure volume relationship was averaged over three cycles after two preconditioning cycles to eliminate hysteresis [11]. Specific compliance of the vessel was determined at each transmural pressure as $C = V^{-1} \Delta V/\Delta P$, where $V$ is the volume at the transmural pressure and $\Delta V/\Delta P$ correspond to increasing increments.

The same procedure was also used to measure the specific compliance of the custom fabricated elastic tubes (described below). As a check of our methodology, the specific compliance of commercially available latex tubes (Dynatek Delta Scientific Instruments, Galena, MO) was also measured ($C = 0.0086 \pm 0.00128 \text{ [1/mmHg]}$) and found to be within approximately 5 percent of the compliance reported by the manufacturer.

Fabrication of Model Compliant Tubing. Custom elastic tubes were fabricated by dipping an aluminum rod 1 in (2.54 cm) diameter and 20 in (50.8 cm) length in a cylinder containing liquid polyurethane solution (Tecoflex Resin SG-85A, Thermedics Inc., Woburn, MA, dissolved in tetrahydrofuran) with viscosity between 760-840 cps. A range of compliance values was achieved by varying the number of dips while allowing 30 min. between dips. Tubes with two dips were found to have comparable compliance as the bovine vena cava and a total of five these tubes were made and tested.

Measurement of Gas Exchange in Compliant and Rigid Tubing. The setup used to measure the gas exchange performance of the respiratory support catheter is shown in Fig. 2. A section of either rigid (Tygon tubing, Cole Palmer, Vernon Hills, IL) or model compliant tubing was mounted in a Plexiglas chamber attached to a flow circuit. Distilled water was pumped at $Q_{\text{water}} = 3 \text{ l/min}$ using a Biomedicus centrifugal pump (Medtronics Inc., Minneapolis, MN) through the vessel test section. An extracorporeal hollow fiber oxygenator (Plexus, Shiley Inc., Irvine,
CA) was placed before the test section to adjust the PO2 and PCO2 of the water to 25–40 mmHg (3.34–5.34 kPa) and 45–50 mmHg (6.67 kPa), respectively. A circulating temperature bath maintained fluid temperature at 37°C. Compliance bags positioned on both ends of the test section (Neoprene breathing bag, Qosina, Edgewood, NY) were used to absorb volume pulsations of the balloon.

Pure oxygen was pulled at Qgas = 3 l/min through the respiratory support catheter fibers under a vacuum pressure of approximately 150 mmHg. Oxygen exchange was determined by measuring the O2 partial pressure (pO2) using an ABL-330 blood-gas analyzer (Radiometer, Copenhagen, Denmark) in samples taken before and after the test section. The O2 exchange was computed according to:

\[ \text{VO}_2 = (p_{O2_{in}} - p_{O2_{out}})(\alpha \times Q_{\text{water}}) \]  

where \( \alpha \) is the O2 solubility in water at 37°C (0.0232 ml/ml/760 Torr). The CO2 exchange rate was determined by measuring the percentage of CO2 in the exhaust gas from the fiber using a mass spectrometer (Medical gas analyzer, Marquette Electronics, Milwaukee, WI). The CO2 exchange was computed using:

\[ \text{VCO}_2 = \% \text{ CO2} \times Q_{\text{gas}} \]

Three sets of measurements were taken at each balloon pulsation rate (0, 30, 60, 120, 180, 240 and 300 bpm). VO2 and VCO2 exchange rates were normalized to fiber surface area, and in addition the VCO2 exchange was also normalized to an inlet pCO2 of 50 mmHg [7]. This minimized the variation in CO2 exchange associated with changes in the inlet pCO2 during and between experiments.

Gas exchange performance was first measured for the rigid tube then the model compliant tube. For the model compliant tube, the transmural pressure was adjusted by changing the chamber pressure until the vessel size yielded comparable CO2 and O2 exchange rates as in the rigid tube without pulsations. This transmural pressure value was maintained at all balloon pulsations. Performance in the model compliant tube and rigid tube was compared by using a Student’s t test, with a value of p<0.05 considered a statistically significant difference (*).

All gas exchange tests used our standard respiratory support catheter (10 series) with a 25 cc balloon, a fiber fabric bundle containing 750 polypropylene hollow fibers (X30-240, Celgard, Charlotte, NC) of 30 cm length, and a total membrane surface area of 0.21 m².

Results and Discussion

Excised Vena Cava and Model Vessel Compliance. Typical pressure-volume curves for excised bovine vena cava and the fabricated polyurethane tube are shown in Figs. 3(a) and 3(b), respectively. The pressure-volume curve for the vena cava specimen (Fig. 3(a) shows a marked decrease in compliance (increase in slope) with transmural pressures above about 6-9 mmHg (0.8–1.2 KPa). Above this transmural pressure, the decreased compliance reflects the increasing stiffness of stretching wall constituents at greater distension or strain, i.e., nonlinear elasticity of wall tissue [11]. Conversely, the fabricated polyurethane tube (Fig. 3(b) generates a negligible transmural pressure until the tube is completely filled (i.e., an essentially infinite compliance until filled), but once filled (31 ml for the segment shown) offers a finite and approximately constant compliance (slope).

The specific compliance, \( C = V^{-1}dV/dP \), associated with all the bovine vena cava specimens (N=5) and fabricated polyurethane tubes (N=5) tested is shown in Figs. 4(a) and 4(b), respectively. The vena cava specific compliance decreases with increasing transmural pressure, consistent with the increased stiffness that the vessel wall exhibits at increasing levels of strain. In our acute animal implants studies [12,13], central venous (i.e., right atrium) and inferior vena cava pressure typically varies from about 5 to 15 mmHg. In this range of transmural pressure the
average specific compliance for all the specimens shown in Fig. 4 is $C=0.0115 \pm 0.00606$ [1/mmHg].

To our knowledge the specific compliance of bovine vena cava has not been previously reported and hence the values measured here cannot be directly compared to other studies. Ohhashi et al. [14] measured the specific compliance of the canine vena cava in the transmural pressure range from 0 to 4 mmHg and reported an average value of $0.19 \pm 0.04$ [1/mmHg] (see Table 1), which is approximately six times the compliance of the bovine vena cava we measured in that same pressure range ($0.0310 \pm 0.014$ [1/mmHg] from Fig. 4). Although species differences are likely, whether this difference can be entirely attributed to species remains unclear. Most studies of vein compliance have addressed veins as arterial vessel autografts [15–18]. Accordingly, these studies have been carried out using femoral arteries [15] or saphenous veins [15–17,19,20], with compliance determination in pressure ranges exceeding 80 mmHg (~10.7 KPa), significantly beyond the transmural pressures considered here. Walden et al. [15] and Abbott et al. [16] have shown that at elevated nonvenous pressures (from 50 to 170 mmHg) (6.7-22.7 KPa) veins have less compliance than arteries, with typical values of $0.220 \times 10^{-2}$ and $0.0295 \times 10^{-2}$ [1/mmHg] respectively (Table 1). In this study, at the maximum transmural pressures load placed on the bovine specimens (about 30 mmHg) (4 KPa) the vena cava compliance was less than 0.005 1/mmHg.

The specific compliance of the fabricated polyurethane tubes once filled (above about 1-2 mmHg) (0.14-0.27 KPa) is essentially constant in our pressure range of interest (~5-15 mmHg). In this range the average compliance for all five fabricated polyurethane tubes is $0.0155 \pm 0.0029$ [1/mmHg] (Table 2). The differences in compliance among the fabricated tubes results from slight variations in wall thickness based on dip casting procedures. For the gas exchange studies described below, we used the longest fabricated tube (symbol $a$ in Fig. 4(b)), which had an average specific compliance of $C=0.0138$ [1/mmHg] over the physiological pressure range. Thus the effective compliance of the model vessel is comparable or greater than vena cava compliance in our pressure range of interest.

### Gas Exchange in Rigid Versus Model Compliant Vessel.

The $O_2$ and $CO_2$ exchange rates for the respiratory support catheter in the rigid (i.e., Tygon tube) versus model compliant vessel (i.e., fabricated polyurethane tube) are compared over a range of balloon pulsation rates from 0 to 300 bpm in Figs. 5(a) and 5(b), respectively. In the absence of balloon pulsation (0 bpm), the $O_2$ (Fig. 5(a)) and $CO_2$ (Fig. 5(b)) exchange rates in the rigid vessel and compliant vessel are comparable, at approximately 12 ml/min/m² for $O_2$ and 21 ml/min/m² for $CO_2$. Vessel size is one of the key determinants of the gas exchange performance of a passive (nonpulsating) intravascular oxygenator [9], since for a given flowrate vessel size dictates the average flow velocity past the oxygenator. Accordingly, that the gas exchange rates in the rigid versus compliant vessel are comparable without balloon pulsation indicates that the average cross-sectional size (i.e., mean diameter or area) of the compliant vessel under the test conditions is effectively equivalent to that of the rigid vessel. Indeed, in our methods we adjusted the transmural pressure in the compliant tube to accomplish this.

Balloon pulsation has a significant and similar effect on the $O_2$ and $CO_2$ exchange rates for both the rigid and model compliant vessels. For $O_2$ (Fig. 5(a), balloon pulsation up to 120 bpm enhanced gas exchange in the model compliant vessel in the same proportion as in the rigid tube, both achieving exchange rates of 67 ml/min/m² at 120 bpm, which represents an approximately 5-fold enhancement relative to no pulsation in both cases. Above

<table>
<thead>
<tr>
<th>Ref.</th>
<th>Vessels</th>
<th>Compliance*</th>
<th>Observation</th>
</tr>
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<tbody>
<tr>
<td>14</td>
<td>Canine femoral vein</td>
<td>0.19 ± 0.04</td>
<td>Compliance measured between 0-3.7 mmHg</td>
</tr>
<tr>
<td>15</td>
<td>Human saphenous vein</td>
<td>0.0220 ± 0.004 x 10²</td>
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<tr>
<td></td>
<td>Human femoral artery</td>
<td>0.0295 ± 0.003 x 10²</td>
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<td>19</td>
<td>Human saphenous vein</td>
<td>0.0225 ± 0.003 x 10²</td>
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<tr>
<td></td>
<td>Canine femoral vein</td>
<td>0.0275 ± 0.003 x 10²</td>
<td></td>
</tr>
<tr>
<td>16</td>
<td>Human saphenous vein</td>
<td>0.0210 ± 0.005 x 10²</td>
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<tr>
<td>17</td>
<td>Human saphenous vein</td>
<td>0.0285 ± 0.005 x 10²</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Human saphenous vein</td>
<td>0.0170 ± 0.002 x 10²</td>
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<tr>
<td></td>
<td>This study</td>
<td>0.0115 ± 0.006</td>
<td>Compliance measured between 5-15 mmHg</td>
</tr>
</tbody>
</table>

* All data had been estimated as $C=V^\prime \Delta P/\Delta V$ [1/mmHg]

### Table 2 Compliant measurements in calves' vena cava and polyurethane tubes over physiological pressure range (5-15 mmHg)

<table>
<thead>
<tr>
<th>Vessels</th>
<th>Compliance mean ± SD [1/mmHg]</th>
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<tr>
<td>Calves' vena cava</td>
<td>0.0112 ± 0.00291</td>
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<tr>
<td>Polyurethane tube</td>
<td>0.0115 ± 0.00727</td>
</tr>
<tr>
<td>Mean ± SD</td>
<td>0.0117 ± 0.00342</td>
</tr>
<tr>
<td>0.0091 ± 0.00830</td>
<td>0.0170 ± 0.00210</td>
</tr>
<tr>
<td>0.0104 ± 0.00721</td>
<td>0.0166 ± 0.00182</td>
</tr>
<tr>
<td>0.0115 ± 0.00607</td>
<td>0.0115 ± 0.00056</td>
</tr>
<tr>
<td>Average</td>
<td>0.0115 ± 0.00806</td>
</tr>
<tr>
<td></td>
<td>0.0155 ± 0.00290</td>
</tr>
</tbody>
</table>

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At 120 bpm, the O₂ exchange rate continually increased with balloon pulsation in the rigid vessel, up to 83 ml/min/m² at 300 bpm, whereas no additional gain in O₂ exchange above 120 bpm was seen for the model compliant vessel. Indeed, above 120 bpm the O₂ exchange rate appears to decrease slightly in the model compliant vessel. The differences in O₂ exchange in the rigid versus model compliant vessel were statistical significant at and above 180 bpm at the p levels indicated.

Similar qualitative behavior is seen for the variation in CO₂ exchange with balloon pulsation in the rigid versus compliant vessel (Fig. 5b). For CO₂, balloon pulsation up to 180 bpm yielded similar gas exchange enhancement in the model compliant vessel as in the rigid tube, both achieving exchange rates of 125 ml/min/m² at 180 bpm, or an approximately 6-fold enhancement relative to no pulsation. Above 180 bpm, the CO₂ exchange rate in the rigid tube continued to increase as balloon pulsation rate increased, up to 131 ml/min/m² at 300 bpm, whereas no additional increase in CO₂ exchange above 120 bpm was seen in the model compliant vessel. As for O₂, the CO₂ exchange rate appears to decrease with balloon pulsation rate above 120 bpm, but only the differences in exchange rate for the rigid versus model compliant vessel at and above 240 bpm were statistically significant at the p levels indicated.

For both O₂ and CO₂ the gas exchange behavior in the model compliant vessel above 120 bpm deviates from the continual increase in gas exchange characteristic of and seen in the rigid vessel. This departure in behavior from the rigid vessel is also seen in the mean pressure drop across the vessel test section, as shown in Fig. 6. Whereas at lower beat rates the mean pressure drop across the compliant vessel test section increases with balloon pulsation, as it does for the rigid vessel over all pulsation rates, above 120 bpm the mean pressure drop across the compliant vessel test section begins to decrease with increasing balloon pulsation. The decreased flow resistance for the compliant vessel above 120 bpm in general terms reflects less interaction of fluid elements traversing the test section with the respiratory support catheter and its fiber surfaces, which imparts drag on these elements. By similar mechanisms, if fluid drag is reduced (i.e., fiber-fluid momentum transfer) mass transfer is reduced, and thus the apparent decrease in gas exchange for both O₂ and CO₂ in the model compliant vessel is at least consistent with the flow dynamics as described by the mean pressure drop across the test section. These linked effects cannot be attributed to a general overall increase in the size of the model compliant vessel above 120 bpm. No marked increase in vessel size was observed, and the mean transmural pressure difference across the model compliant vessel (dashed line in Fig. 6, right ordinate) remains nearly constant or decreases slightly with increasing balloon pulsation rate.

The genesis of the decreased gas exchange and mean pressure drop above 120 bpm can be related to substantial local motion of the vessel wall immediately below (inferior to) the pulsating balloon, a motion that intensified at the higher beat rates. Much of the observable effect of balloon pulsation on the model compliant vessel was seen as movement of the immediately inferior wall to and from the fiber bundle and pulsating balloon. While this motion was seen at all pulsation rates and increased with increasing pulsation rate, the inferior wall motion was noticeably stronger above 120 bpm. Figure 7 shows images of the compliant vessel at balloon deflation during no pulsation (0 bpm, Fig. 7(a)) and pulsation at 60 bpm (Fig. 7(b)) and 240 bpm (Fig. 7(c)). The model compliant vessel is approximately cylindrical with no pulsation and shows only slight wall movement with pulsation at 60 bpm (b). Conversely, the displacement of the inferior wall at a balloon pulsation rate of 240 bpm (c) is appreciable. Furthermore, above 120 bpm the inferior wall noticeably flattens during balloon pulsation (not shown), appearing almost “floppy” during part of the pulsation cycle. This indicates that during balloon deflation at the higher pulsation rates the local transmural pressure difference at the vessel wall below the balloon is near zero, i.e., is sufficiently small that the wall is no longer in tension. Unlike the
natural vena cava, the compliance of the model compliant vessel rapidly increases near zero transmural pressure (see Fig. 4(b)) and motion of the vessel wall would be intensified by this large compliance.

Thus our conclusion is that the unnaturally large compliance of our fabricated compliant vessel at small transmural pressures leads to local changes in vessel geometry at the higher pulsation rates that reduces the interaction of traversing fluid with the fiber bundle, thus reducing gas exchange and the mean pressure drop as a result. If our custom fabricated tubing had shown finite compliance at small transmural pressure, as does the normal vena cava, we would have expected that the vessel wall motion underneath the pulsating balloon would not have been accentuated at the higher beat rates, as seen in these studies. In the absence of that accentuated wall motion the O₂ and CO₂ exchange rates in the compliant model vessel should be similar to that for the rigid vessel at all balloon pulsation rates. In support of this conclusion, our acute in-vivo studies of the respiratory support catheter have not indicated that a decrease in gas exchange or blood pressure drop occurs as balloon pulsation rate increases above a particular value. We also have not seen accentuated wall motion of the vena cava at higher pulsation rates when we have observed the respiratory catheter pulsing in-situ during open-chested animal preparations.

In this study we chose not to investigate the performance of the respiratory catheter at various levels of vessel compliance. The compliance of our model compliant vessel was close to the high end of compliance measured for the bovine vena cava specimens. Had we found, using this model compliant vessel, that vessel compliance per se does influence the gas exchange performance of the respiratory support catheter we would have needed to assess the effects of compliance over a broader range, including compliance values intermediate to those investigated here.

Our bench studies of respiratory catheter performance are not intended to mimic in-vivo conditions or predict how a given catheter prototype will specifically perform in-vivo. The in-vivo environment of the respiratory catheter is complicated by features that are difficult to model in-vitro, including irregular shape and size of the vena cava, distributed veins draining into the vena cava and the compliance that these veins provide for flow and pressure pulsations, bidirectional flow past the catheter because of placement partly in the superior vena cava and partly in the inferior vena cava, and potential physiological variations in vessel geometry and flow with time. The bench test of the respiratory catheter is designed primarily as a tool to characterize and compare the performance of different prototype designs along the development pathway. In these bench characterization tests we perfuse with distilled water rather than blood for simplicity and to accommodate the need for multiple and repeatable tests on the same device. Our experience is that catheter performance in water is a good relative predictor of in-vitro performance in blood (i.e., a catheter design yielding greater gas exchange performance in water will also show superior performance in blood). Vaslef et al. have shown a similar correspondence of the gas exchange levels in blood versus water using commercially available membrane oxygenators. Our desire to address the effects of vessel compliance in the bench test is driven less by wanting a more realistic test setup of venous conditions than by recognition that comparative evaluation of balloon enhanced gas exchange among different catheter prototypes might be affected if vessel wall motion resulting from vessel compliance is important. Our conclusion based on these studies is that vessel compliance per se is most likely not significant.
important and that bench tests in rigid model vessels should offer a suitable platform for comparative evaluation of pulsating respiratory catheter prototypes.

Summary and Conclusions

The specific compliance of the bovine vena cava decreased with increasing pressure. In a physiological relevant pressure range (5-15 mmHg), the average specific compliance of all five specimens tested was 0.0115±0.00606 [1/mmHg]. The specific compliance of the fabricated polyurethane tube was approximately constant in the same pressure range with average value of 0.0155±0.0029 [1/mmHg].

Gas exchange levels in the rigid and model compliant tubes increased with 0 balloon pulsation at the same rate up to a pulsation frequency of 120 bpm. Above this frequency no increase in gas exchange occurred in the model compliant vessel, whereas gas exchange in the rigid vessel continued to increase. We attribute this behavior to the local movement above 120 bpm of the vessel wall immediately below the pulsating balloon. This behavior is most likely not seen in physiological application of the respiratory support catheter because unlike the natural vena cava, the model compliant tube has a large (effectively infinite) compliance near zero transmural pressure.

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