A Model of Venous Return While Utilizing Vacuum Assist During Cardiopulmonary Bypass

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Abstract: In order for vacuum-assisted venous return (VAVR) to be used safely and efficiently during cardiopulmonary bypass (CPB), a full understanding of venous return is necessary. The focus of this work was to use the concepts of energy conservation and viscous energy dissipation in the development of a theoretical model of venous return utilizing vacuum assist. The effectiveness and accuracy of this model has been verified through in vitro laboratory investigations and statistical analysis. Although VAVR can provide higher flows through smaller venous cannula, vacuum assist may lead to increased levels of wall shear stress as shown in this work. The clinical implications of VAVR have yet to be investigated, but may lead to an exacerbation of the detrimental effects of CPB during cardiac surgery. Keywords: venous return, vacuum assist, modeling, shear stress, flow. JECT 2003;35:224–229

Traditionally, venous return to the cardiotomy reservoir on the heart-lung machine has been made possible by creating a siphon (gravity drainage) in the venous line(s). Recently, the introduction of smaller, less cumbersome venous cannulas has limited the amount of venous return that can be achieved with traditional siphon drainage (1,2). This limitation challenges the maintenance of adequate systemic flows; therefore, the utilization of assisted venous drainage techniques is again becoming common practice. Vacuum-assisted venous return (VAVR) is one form of assisted venous drainage accomplished by connecting a vacuum source to the cardiotomy reservoir. This technique increases the pressure drop across the venous cannula and lines, thus augmenting venous return (3,4). With VAVR, venous return rate becomes a function of both the siphon effect and the amount of negative pressure (vacuum) applied.

To date, the efficacy of using VAVR has not been fully investigated outside of its utility in achieving higher flow rates through smaller bore venous cannula and lines (5,6). Previous studies have been centered on in vitro and in vivo measurements done with little or no theoretical analysis. Also, no models of venous return have been developed that consider the pressure drops and shear forces associated with different venous line configurations and venous cannulas. VAVR does allow for the miniaturization of the perfusion circuitry and reduced priming volumes; however, on a clinical level, its implications have yet to be thoroughly investigated.

The focus of this research was to develop valid, theoretical models of venous return from which the effects of VAVR utilization could be determined. These models can predict pressures, flows, resistances, and wall shear stresses at any physical location within a venous return circuit. Theoretical models were developed for both the utilization of a single, dual-stage venous cannula (atrial-caval) and two single-stage (bi-caval) venous cannulas. An in vitro laboratory investigation was used to verify the validity of the proposed models. The goal was to develop a tool that perfusionists and surgeons could use to make informed decisions regarding the application of VAVR to cardiopulmonary bypass.

MATERIALS AND METHODS

Theoretical Models That Predict Pressures, Flows, and Resistances at Various Points in a Venous Return Circuit

The models were derived using the energy balance equation for fluid systems (Bernoulli Equation). The terms of this equation describe the various forms of energy at any given point in a fluid system and can be used to determine the interconversions as fluid moves from one...
section of the system to another (7–9). Specifically, it can account for energy losses (changes) attributable to viscous (frictional) effects as well as changes attributable to conduit diameter, convergences, divergences, inlet conditions, outlet conditions, and bends.

To apply the energy balance model to the venous return system, the following assumptions were made. First, the blood was considered to be a Newtonian, incompressible fluid. Second, it was assumed that the flow was fully developed at all times, thus neglecting entrance lengths. Last, the system was assumed to be rigid and closed. This meant that the net flow through the system must be constant.

The two theoretical models developed were for the cases of utilizing a single dual-stage venous cannula (atrial–caval) and two single-stage (bi-caval) venous cannulas. The circuits used to verify the theoretical models are shown in Figures 1 and 2.
Model for the Atrial–Caval Cannulation Technique

\[ P_1 = P_c + \gamma \left( Z_c - Z_1 - \frac{V_1^2}{2g} + h_{f,3,4} + h_{\text{exit}} + h_{\text{div, flow}} \right) + 2'h_{f,2,3} + h_{\text{con, flow}} + h_{\text{entrance}} + h_{f/1.2} \]  
\[ P_2 = P_c + \gamma \left( Z_c - Z_2 - \frac{V_2^2}{2g} + h_{f,3,4} + h_{\text{exit}} + h_{\text{div, flow}} \right) + 2'h_{f,2,3} + h_{\text{con, flow}} \]  
\[ P_3 = P_c + \gamma \left( Z_c - Z_3 - \frac{V_3^2}{2g} + h_{f,3,4} + h_{\text{exit}} \right) \]  
\[ P_4 = P_c + \gamma Z_c \]  

Where: \( P_1 \) = pressure at point 1, \( P_2 \) = pressure at point 2, \( P_3 \) = pressure at point 3, \( P_4 \) = pressure at point 4, \( P_c \) = cardiotomy pressure, \( Z_1 \) = height of point 1 above the reference, \( Z_2 \) = height of point 2 above the reference, \( Z_3 \) = height of the fluid in the cardiotomy above the reference, \( V_1 \) = fluid velocity at point 1, \( V_2 \) = fluid velocity at point 2, \( V_3 \) = fluid velocity at point 3, \( \gamma \) = specific weight, \( g \) = gravitational constant, \( h_{f,1.2} \) = frictional loss between points 1 and 2, \( h_{f,2,3} \) = frictional head loss between points 2 and 3, \( h_{f,3,4} \) = frictional head loss between points 3 and 4, \( h_{\text{exit}} \) = head loss attributable to exit effects, \( h_{\text{div, flow}} \) = head loss due to the divergence of flow, \( h_{\text{con, flow}} \) = head loss attributable to convergence of flow, \( h_{\text{entrance}} \) = head loss attributable to entrance effects.

Model for the Bi-Caval Cannulation Technique

\[ P_{\text{SVC}} = P_c + \gamma \left( Z_c - Z_{\text{SVC}} - \frac{V_{\text{SVC}}^2}{2g} + h_{f,4,5} + h_{\text{exit}} + h_{f/\text{SVC}} \right) + h_{\text{con, 3,4}} + h_{\text{ent, SVC}} + h_{f/\text{SVC, canna}} \]  
\[ P_{\text{IVC}} = P_c + \gamma \left( Z_c - Z_{\text{IVC}} - \frac{V_{\text{IVC}}^2}{2g} + h_{f,4,5} + h_{\text{exit}} + h_{f/\text{IVC}} \right) + h_{\text{con, 2,4}} + h_{\text{ent, IVC}} + h_{f/\text{IVC, canna}} \]  
\[ P_2 = P_c + \gamma \left( Z_c - Z_2 - \frac{V_2^2}{2g} + h_{f,4,5} + h_{\text{exit}} + h_{f/\text{SVC}} \right) + h_{\text{con, flow, 2,4}} \]  
\[ P_3 = P_c + \gamma \left( Z_c - Z_3 - \frac{V_3^2}{2g} + h_{f,4,5} + h_{\text{exit}} + h_{f/\text{IVC}} \right) + h_{\text{con, flow, 3,4}} \]  
\[ P_4 = P_c + \gamma \left( Z_c - Z_4 - \frac{V_4^2}{2g} + h_{f,4,5} + h_{\text{exit}} \right) \]  
\[ P_5 = P_c + \gamma Z_c \]  

Where: \( P_{\text{SVC}} \) = pressure in the superior vena cava (SVC), \( P_{\text{IVC}} \) = pressure in the inferior vena cava (IVC), \( P_2 \) = pressure at point 2, \( P_3 \) = pressure at point 3, \( P_4 \) = pressure at point 4, \( P_5 \) = pressure at point 5, \( P_c \) = cardiotomy pressure, \( Z_{\text{SVC}} \) = height of the SVC above the reference, \( Z_{\text{IVC}} \) = height of the IVC above the reference, \( Z_2 \) = height of point 2 above the reference, \( Z_3 \) = height of point 3 above the reference, \( Z_4 \) = height of point 4 above the reference, \( Z_c \) = height of the fluid in the cardiotomy above the reference, \( V_{\text{SVC}} \) = fluid velocity in the SVC, \( V_{\text{IVC}} \) = fluid velocity in the IVC, \( V_2 \) = fluid velocity at point 2, \( V_3 \) = fluid velocity at point 3, \( V_4 \) = fluid velocity at point 4, \( \gamma \) = specific weight, \( g \) = gravitational constant, \( h_{f/\text{SVC, canna}} \) = frictional head loss in the SVC cannula, \( h_{f/\text{IVC, canna}} \) = frictional head loss in the IVC cannula, \( h_{\text{ent, SVC}} \) = frictional head loss in the SVC line, \( h_{\text{ent, IVC}} \) = frictional head loss between points 4 and 5, \( h_{\text{exit}} \) = head loss attributable to exit effects, \( h_{\text{con, 2,4}} \) = head loss attributable to the convergence of flow from points 2 and 4, \( h_{\text{con, 3,4}} \) = head loss attributable to the convergence of flow from points 3 and 4, \( h_{\text{ent, SVC}} \) = head loss attributable to entrance effects into the SVC cannula, \( h_{\text{ent, IVC}} \) = head loss attributable to entrance effects into the IVC cannula.

In each equation, the pressure in the cardiotomy (\( P_c \)) is zero when the cardiotomy is vented to the atmosphere. When vacuum assist is used, it is equal to the amount of vacuum applied. The energy losses attributable to geometric changes and frictional effects were accounted for by the head loss terms (\( h_{\gamma} \)).

The application of the models to the venous return circuit was accomplished with the use of a spreadsheet to solve Equations 1 through 4 and 5 through 10 simultaneously. The assumption was made that if the patient was optimally drained, the pressures in the right atrium (\( P_1 \)), superior vena cava (\( P_{\text{SVC}} \)), and inferior vena cava (\( P_{\text{IVC}} \)) would approach zero. By forcing these values to be zero and giving the physical dimensions of the circuit components, it was possible to solve for flow.

Theoretical Model Predicting Wall Shear Stress at Various Points in a Venous Return Circuit

Shear stress is a function of the viscosity and the shear rate of the fluid flowing through a network of tubing (10). Viscosity is the internal friction or resistance of fluid to flow; whereas, shear rate is the rate of velocity reduction in the direction perpendicular to the path of fluid flow (9). The value of shear stress becomes a maximum when the distance from the center of the tube, \( r \), is equal to the tube radius, \( R \) (9,11). Therefore, the maximum shear stress is
located at the wall of the tube and is calculated using Equation 11 (8,11).

\[ \tau_{\text{max.}} = \tau_{\text{wall}} = \frac{\Delta P \cdot R}{2L} \]  

(11)

Where: \( \Delta P \) = pressure drop along the tube between two points described by Equations 1–10, \( R \) = radius of the tube, and \( L \) = length of the tube.

If the pressure drop across any section of tubing is known along with the radius and the length of the tube, the wall shear stress can be estimated. Because the pressure drop in the venous circuit is typically greatest across the venous cannula, the magnitude of wall shear stress in this component was a focus of this investigation.

**Verification Circuit Equipment**

The circuits used to verify the theoretical models of venous return (shown in Figures 1 and 2) were constructed using a Bio-Medicus 550 Centrifugal Pump (Medtronic, Minneapolis, MN), Bard CPS heat exchanger (C.R. Bard, Murray Hill, NJ), Avecor Cardiotomy/Venous Reservoir (Medtronic, Minneapolis, MN), and Terumo Hard Shell Venous Reservoir (Ann Arbor, MI). The right atrium was simulated with an Avecor Soft Shell Venous Reservoir (Medtronic, Minneapolis, MN). The vacuum source was a Medela Low Vacuum Aspirator (McHenry, IL). Blood gas values were obtained using the IRMA SL Blood Analysis System (Agilent Technologies, Palo Alto, CA). All pressure values were recorded using a Biotek DPM-2 Plus Universal Pressure Meter (Winooski, VT).

The cannulas used in the laboratory investigation included the following: DLP 24 Fr. model 66124 (Medtronic, Minneapolis, MN), DLP 18 Fr. model 66118, DLP 12 Fr. model 66112, DLP 29/29 Fr. model 91329, Sarns 28/38 Fr. (Medtronic, Minneapolis, MN), and Terumo Hard Shell Venous Reservoir (Ann Arbor, MI). The right atrium was simulated with an Avecor Soft Shell Venous Reservoir (Medtronic, Minneapolis, MN), and Terumo Hard Shell Venous Reservoir (Ann Arbor, MI). The left atrium was simulated using a Bio-Medicus 550 Centrifugal Pump (Medtronic, Minneapolis, MN). The vacuum source was a Medela Low Vacuum Aspirator (McHenry, IL). Blood gas values were obtained using the IRMA SL Blood Analysis System (Agilent Technologies, Palo Alto, CA). All pressure values were recorded using a Biotek DPM-2 Plus Universal Pressure Meter (Winooski, VT).

**Laboratory Procedure**

To verify the accuracy of the theoretical models, data were collected using the test circuits as shown in Figures 1 and 2. A randomized, repeated measures design was used in the experimental methodology.

The test circuits were primed with a mixture of heparinized bovine blood and Plasmalyte-A™ to yield a hematocrit of 21%. The range of cardiotomy (vacuum) pressures used was 0 to −80 mmHg in increments of −10 mmHg. For each level of vacuum, flow and pressure measurements were taken at each specified point (see Figures 1 and 2). Blood gas samples were taken at the beginning and end of each set of trials. A total of five trials were performed for each cannula.

**Statistical Analysis**

The ability of the models to predict flow and wall shear stress accurately was determined using regression analysis and \( t \)-statistic calculation (12). Two-way analysis of variance (ANOVA) and the Bonferroni \( t \)-test were used to evaluate the differences between the amount of vacuum applied and the corresponding flow and wall shear stress values. Statistical significance was determined to be at a \( P \) value less than .05 (\( P < .05 \)).

**RESULTS**

Regression analysis of the measured and predicted flows yielded the regression equation shown in Equation 12. Calculation of the \( t \)-statistic verified that the slope of the regression line was not different from 1 (\( t = −0.722, t_{\text{critical}} = 1.645, \alpha = 0.05, \text{df} = 224 \)), indicating that the model is an exact predictor of flow over the range of vacuum specified (12) (Figure 3).

Measured flow = 0.0057 + 0.9977 * Predicted flow  

(12)

ANOVA analysis in combination with the Bonferroni \( t \)-test verified that there was a statistically significant change in flow for each incremental (10 mmHg) change in vacuum applied to the cardiotomy reservoir (\( P < .01 \)).

Referring to Figure 4, the predicted wall shear stress values are calculated based on the mathematical models, while the estimated wall shear stress values are calculated utilizing Equation 11 and the pressure drops measured in the test circuits. Our lab did not have the ability to measure wall shear stress values directly. Regression analysis of the predicted shear stress data with the estimated data produced the regression equation shown in Equation 13.

Estimated wall shear stress = 36.390  

\[ + 0.5569 \cdot \text{Predicted wall shear stress} \]  

(13)

Calculation of the \( t \)-statistic indicated that the slope of the line is significantly different from 1 (i.e., perfect correlation) (\( t = −17.013, t_{\text{critical}} = 1.645, \alpha = 0.05, \text{df} = 224 \) (12). This shows that the shear stress model is not a perfect predictor of cannula wall shear stress over the range of vacuum tested. When the \( t \)-statistic was used to test the slope of the regression equation to zero (i.e., no correlation) (12), it was shown that there was a positive correlation between predicted and estimated cannula wall shear stress (Figure 4) (\( t = 21.38, t_{\text{critical}} = 1.645, \alpha = 0.05, \text{df} = 224 \)).

ANOVA analysis in combination with the Bonferroni \( t \)-test verified that there was a statistical significant change in cannula wall shear stress for each incremental (10 mmHg) change in vacuum applied to the cardiotomy reservoir (\( P < .01 \)).

**DISCUSSION**

The goal of the present work was to develop a model that could be used to predict pressure, flow, resistance, and wall shear stress values at various physical locations in a venous circuit over a range of cardiotomy pressures,
given the venous line configuration and cannula size. From the statistical analysis performed on the data collected in the laboratory investigation and obtained from the theoretical model, it can be concluded that the theoretical model is an exact predictor of flow over the range of vacuum specified. Although not a perfect predictor, the theoretical model derived to predict wall shear stress was shown to have positive correlation to the cannula wall shear stress over the same range of vacuum pressures.

As with any model, there are limitations to their applicability. Assumptions need to be made to apply some of the fundamental equations used in this work. When modeling the flow of blood in tubes, common assumptions used are that blood is a Newtonian fluid and that the viscosity remains constant, neither of which is the case. Blood is a non-Newtonian fluid with a viscosity that varies inversely with the shear strain rate, better known as the apparent viscosity (10). Considering this and the fact that the theoretical model for cannula wall shear stress did not show perfect correlation, future work should concentrate on taking into account the complex nature of blood and its flow dynamics. However, despite the assumptions, this model was predictive of wall shear stress.

With respect to the current application of vacuum-assisted venous return, the ability to model a given venous line configuration and obtain feedback as to theoretically...
feasible flow rates and estimated wall shear stress values has not yet been proposed. The models developed in this work can provide this useful information to the perfusionist and surgeon, giving them the ability to determine safe limits of VAVR and its clinical application.

Several researchers have investigated the importance of shear stress when looking at the flow of blood through tubes. Previous studies have indicated that high levels of shear can activate leukocytes and platelets as well as result in the destruction of red blood cells (hemolysis). Leukocyte activation occurs at wall shear stress levels between 75 to 800 dynes/cm² (13). Similarly, platelet activation has been shown to occur at wall shear stress levels between 60 and 100 dynes/cm² (13–15). Finally, wall shear stress values between 1500 and 3000 dynes/cm² have been shown to cause measurable amounts of hemolysis (13,16). All of these factors can impair hemostasis and possibly contribute to the post-operative morbidity associated with cardiopulmonary bypass.

Vacuum-assisted venous return has proved to be an effective tool in the clinical setting because it enables the use of smaller venous lines and cannulas while being able to maintain adequate systemic flows. This has helped to reduce prime volumes as well as allow the surgeon to use cannulas that are less obstructive and cumbersome. Even though the model did not perfectly predict wall shear stress values, both the model and the lab results show that the use of VAVR significantly increases cannula wall shear stress (Figure 4). Based on these high levels of shear stress, there is potential for detrimental effects associated with the use of VAVR. The magnitude of the increase in wall shear stress shown in this work warrants further investigation on this topic as it pertains to the use of VAVR during cardiopulmonary bypass.

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