Intravenous Access and Delivery Principles

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Summary

Intravenous (IV) administration of fluids and drugs is an important part of the routine care of patients, especially those undergoing anesthesia and surgery. The IV infusion is essential as a route of drug administration. In addition to facilitating induction, IV drugs are administered throughout the surgical procedure both as part of the anesthetic technique and as a way to meet surgical needs. Whenever fluid or blood is lost during surgery, replacement must be considered, sometimes with great immediacy and sometimes electively. Also, the patient's insensible losses over and above those from surgery must be replaced.

Every year in the United States, at least 20 million patients receive IV fluids. Although clinicians generally regard IV fluid administration as safe and effective, IV site complications occur frequently. These include phlebitis, thrombosis, catheter obstruction, and fluid extravasation. From the anesthesiologist's standpoint, the obvious result of these complications is inability to administer the intended drugs and fluids to the patient.

IV systems used in anesthesia range from the most simple to the most complex. Drugs can be administered from a syringe with needle into the flowing IV stream that is powered by gravity. At the other extreme, pumps or controllers can create the continuous infusion, and a syringe or other pumps can administer drugs. These drugs can be controlled manually by the clinician or by a computer, which determines the patient's drug needs based on a pharmacokinetic model with or without the assistance from the patient's physiologic variables.

This chapter summarizes alternative drug and fluid administration techniques and devices in the framework of a simple physical, physiologic model so that one can understand fluid flow in IV systems and patient veins.

HISTORY

Until the 1970s, IV infusion was performed with simple systems. An elevated, fluid-filled bottle provided the energy while the clinician (nurse or physician) adjusted a roller clamp until the correct rate of drop formation was observed in the plastic drip chamber. As clinicians questioned the limitations of their tools, they discovered certain inadequacies in
the conventional approach. In 1966 La Cour and Ferechak et al. in 1971 showed that the rate of drop formation is not an accurate measure of flow rate because of the influence of temperature, fluid composition, and diameter and shape of the drip chamber orifice. Flack and Whyte showed that cold flow (creep) in the tubing underlying the roller clamp contributes to flow variation, and Ziser et al. found this change to be greater than 15% over 45 minutes.

Venous physiology has been studied much more extensively than the technology of fluid infusion. Special focus has been placed on vein collapse. A review of that literature explains our current knowledge. In 1912 Starling (with Knowlton) used a collapsible thin-walled rubber tube to produce constant back pressure (afterload) in his studies of heart failure. Although he claimed control of resistance externally, wall flutter was seen and found that collapsible thin-walled rubber tube to produce conditions of tube collapse, flow is independent of the resistance and the resulting decreased venous flow, attributing it to changing resistance. In 1954 Duomarco and Rimini demonstrated the importance of energy and hydraulic gradients along veins. Rodbard and Saiki found that "flow through widely patent elastic tubes followed the laws for flow through rigid pipes." They also noted that when air compressed a thin-walled venous model externally, wall flutter was seen and heard. Flutter ceased when external compressing pressure was raised.

Rodbard showed that critically high flow velocities occur when veins collapse and explained the paradoxic increase in flow from vessel distension produced by adding downstream resistance. He showed later that constant flow in various body tissues can be produced by the same collapse phenomenon.

In 1963 Permutt and Riley explained that under conditions of tube collapse, flow is independent of the pressure drop from inlet to outlet. Rather, flow "depends on the difference between inlet pressure and critical closing pressure" (i.e., occlusion pressure or external pressure, \( P_e \)). They used the term vascular waterfall to describe this phenomenon, applied by Starling in 1912.

In 1969 Holt and Conrad analyzed the physics of the collapse process observed in rubber tubes. Conrad's photos of a partly collapsed tube have been reprinted many times in texts and treatises that summarize the literature. Later, Kresch and Noordergraaf performed additional analyses of collapsible rubber tubes, whereas Katz et al. found that "the significant variable is transmural pressure." Griffiths noted that when collapse occurs, fluid flows at sonic velocity (i.e., the velocity of pressure waves in the tube). Brower and Noordergraaf then showed that "negative resistance" can explain the results of collapsed-tube studies. Further analysis by Shapiro developed the one-dimensional theory of steady flow in thin-walled tubes partly collapsed by negative transmural pressure. Guyton has written extensively on both vein and tissue pressures. The compliance of collapsible vessels has also been studied.

The models just described explain the behavior of thin-walled rubber tubes subjected to external pressure that exceeds internal pressure. The incomplete collapse of rubber tubes makes the analysis complex. Ironically, in vivo and in vitro veins probably do not behave in this complex way; they collapse completely (unpublished data, D. Joseph and the author). Inapplicable analysis has confused our understanding of the behavior of clinical IV systems.

**SIMPLE IV SYSTEM**

The simplest IV system consists of an elevated fluid-filled bag or bottle, a length of flexible tubing, a catheter cannulating the patient's vein, and the patient's venous system, which ultimately terminates at the heart, specifically the right atrium. An adjustable roller clamp compresses the tubing at one point, slowing fluid flow and allowing manual control by a clinician. The rate at which drops form is assessed visually, and the roller clamp is adjusted manually as needed. Depending on the type and brand of the tubing system, drop size is 1/10, 1/15 ml (macrodrop), or 1/60 ml (microdrop).

The flow rate through the IV system depends on several factors. The height of the IV bag less the pressure in the patient's vein determines the driving pressure for flow. The resistance of the catheter plus tubing system then determines flow:

\[ F = \Delta P / R \]

where \( F \) is flow, \( \Delta P \) is pressure difference, and \( R \) is resistance. In this analysis, \( F \) and \( \Delta P \) are generally measured and \( R \) is computed.

The pressure difference depends on both fluid height in the bag or bottle and pressure in the patient's vein. The latter is more complex than is usually realized.

Veins possess two characteristics that slow fluid flow: resistance to fluid flow and opening pressure. Resistance is caused by the long, narrow nature of veins, which causes pressure loss along their length. Venous blood flow is usually assumed to be laminar, a subject discussed later in this chapter. Opening pressure for the vein depends on the forces exerted by tissues outside the vein along its course.

Before fluid is forced through the vein, venous pressure is essentially indeterminate. That is, the
pressure measured may take on any value less than the obstructing or external pressure. Once fluid flows through the vein, pressure rises to the opening pressure. Then venous resistance affects fluid flow, and pressure rises as flow increases. These concepts and their application are developed further throughout this chapter.

**PRESSURE-FLOW RELATIONSHIPS AND MODERN THEORY**

To understand fluid flow in IV systems, one can use a conceptual model. The model describes the fluid flow behavior of any IV system and patient. Actual components of the system are depicted as combinations of one or more ideal components. The behavior of actual components can be predicted and measured using the model along with simple measurements.

The system is comprised of a selection of components from Table 55-1. Fig. 55-1 shows the theoretic pressure-flow relationship (PFR) for each component. The simplest configuration of actual components, a gravity flow administration system with rate manually controlled by a roller clamp, is shown in Fig. 55-2. Pressures at various locations, depicted as the heights of the vertical bars above the reference level, are shown under the condition that flow is 200 ml/hour.

The model's behavior can be viewed as follows. Pressure at the free-air surface at the top of the bag or bottle is zero (gauge), that is, atmospheric pressure. Lower in the tubing, the pressure head proximal to the roller clamp is determined by the height of the free-air surface above the roller clamp, combined with fluid density and gravity. Specifically, pressure equals height times fluid density times the acceleration of gravity. When the liquid is water and pressure is expressed in centimeters of water (cm H₂O), pressure is equal to the water height in centimeters. The density of most clinical solutions differs little from one (1.00).

Because an air-filled drip chamber intervenes between bag and roller clamp, the head height is diminished by the height of the air column. The diminution of pressure occurs because within the air-filled chamber, pressure is the same everywhere. No vertical pressure gradient exists in the air-filled chamber. The tubing is drawn vertically from the

<table>
<thead>
<tr>
<th>Actual component</th>
<th>Ideal component</th>
<th>Symbol</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bag</td>
<td>Pressure source</td>
<td>( P_{\text{bag}} )</td>
</tr>
<tr>
<td>DS roller clamp</td>
<td>Variable resistor</td>
<td>( R_{\text{clamp}} )</td>
</tr>
<tr>
<td>Tubing</td>
<td>Low-resistance conduit</td>
<td>( R_{\text{tub}} )</td>
</tr>
<tr>
<td>Catheter</td>
<td>Resistor</td>
<td>( R_{\text{cath}} )</td>
</tr>
<tr>
<td>Normal peripheral vein</td>
<td>Resistor</td>
<td>( R_{\text{vln}} )</td>
</tr>
<tr>
<td>Extravascular tissue</td>
<td>Resistor</td>
<td>( R_{\text{vln}} )</td>
</tr>
<tr>
<td>Central vein</td>
<td>Conduit with no resistance</td>
<td>( R_{\text{vln}} = 0 )</td>
</tr>
<tr>
<td>Heart</td>
<td>Pressure source</td>
<td>( P_{\text{cv}} ) = central venous pressure</td>
</tr>
<tr>
<td>Obstruction</td>
<td>Starling resistor = Collapsible tube</td>
<td>( P_{\text{ext}} )</td>
</tr>
<tr>
<td>Inflated blood pressure cuff</td>
<td>Collapsible tube (CT)</td>
<td>( P_{\text{ext}}, R_{\text{vln}} )</td>
</tr>
<tr>
<td>Venous tourniquet</td>
<td>CT</td>
<td>( P_{\text{ext}}, R_{\text{vln}} )</td>
</tr>
<tr>
<td>Collapsible tube</td>
<td>CT</td>
<td>( P_{\text{ext}}, R_{\text{tub}} )</td>
</tr>
<tr>
<td>Problem IV site</td>
<td>CT</td>
<td>( P_{\text{ext}}, R_{\text{vln}} )</td>
</tr>
<tr>
<td>Catheter against vein wall</td>
<td>CT</td>
<td>( P_{\text{ext}}, R_{\text{vln}} )</td>
</tr>
<tr>
<td>Volumetric pump</td>
<td>Flow source</td>
<td>( F_{\text{pump}} )</td>
</tr>
<tr>
<td>Volumetric controller</td>
<td>Variable resistor</td>
<td>( R_{\text{cont}} )</td>
</tr>
<tr>
<td></td>
<td>Variable pressure source</td>
<td>( P_{\text{effective}} )</td>
</tr>
<tr>
<td>Observation of drop rate</td>
<td>Flow measure</td>
<td>( F_{\text{drop}} )</td>
</tr>
<tr>
<td>Electronic pressure transducer</td>
<td>Pressure sensor</td>
<td>( P )</td>
</tr>
<tr>
<td>Water manometer</td>
<td>Pressure sensor</td>
<td>( P )</td>
</tr>
</tbody>
</table>

*Actual components behave similarly to theoretic components.*
bottle outlet and then horizontal until entering the patient's vein, which is at the same height as the heart (right atrium). This simplifies the physics to make the system more understandable.

In Fig. 55-2 the roller clamp has been adjusted to a resistance that provides a flow rate of 200 ml/hour. In clinical practice the clinician adjusting the clamp is unaware of actual resistance. He/she simply rotates the roller clamp until proper flow rate is judged by measuring the rate of drop formation observed in the drip chamber. Here, pressure just distal to the roller clamp is measured to be 16 cm H₂O with this particular combination of system components and patient. From the values for pressure difference across the clamp and the value for flow, roller clamp resistance can be computed.

Next, the origin of the pressure distal to the roller clamp must be considered. From the standpoint of the tubing system, the pressure at the catheter tip originates in the patient. The peripheral venous pressure measured in this situation can best be understood by realizing that the resistance of the vein cannot be zero. Thus the flow of blood back to the heart makes peripheral venous pressure slightly higher than central venous pressure (CVP), in the absence of fluid infusion and intervening obstruction. The pressure increment often is quite small (3 cm H₂O or less). With partial venous obstruction (see later discussion) venous pressure may be significantly greater.

PRESSURE-FLOW RELATIONSHIP DATA

The PFR for the tubing-catheter-patient system can be measured in experimental situations and in clinical practice. This can be done with a volumetric pump and a pressure-monitoring device such as a pressure transducer or even a water column. One particular commercial IV infusion pump can facilitate this. The IVAC variable-pressure volumetric infusion pump, Model 560 (IVAC Corp., San Diego), produces constant flow and measures pressure during infusion and when the pump is stopped. In this pump, flow is created by a linear peristaltic mechanism that provides a flow pattern composed of successive 1 µl volumes. Just distal to the pumping mechanism, a pressure transducer presses against an in-line pressure-sensing disk contained in the IV tubing. The pressure-monitoring pump measures pressure with an accuracy of ± 2 mm Hg. Pressure is measured in both the presence and the absence of flow.

Using the pump, the PFR has been measured for many tubing systems, catheters, other devices, and patients. Because of flow limitations in the pump (1000 ml/hour = 1 L/hour), effects of high flow on system nonlinearity cannot be studied with this device. High flow effects are presented in the section on nonlinearity. Resistance units (RU) are mm Hg/L/hour.

From experiments using the pump, the following facts have been determined. An elevated IV bag behaves as a pressure source, with pressure equal to
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CHAPTER 55

Fig. 55-2. Model for manually controlled IV administration system and patient. A, Pictorial schematic diagram of model's components. B, Pressures (cm H₂O) observed in various locations. C, Flow chart showing pressures (circles), resistances (rectangles), collapsible tubes (triangles), and pressure drops (arrows). In A, elevated fluid-filled bag acts as pressure source (P = 100 cm H₂O in B) equal to height of its free-air surface above reference level. Roller clamp is resistor, arbitrarily adjusted for F (flow) = 200 ml/hour in A. Catheter provides additive resistance in fluid path to patient. Vein acts as hydraulic resistor as well but can collapse if external pressure (8 cm H₂O in B) exceeds internal pressure (16 cm H₂O in B). Collapse does not occur here because internal pressure exceeds external pressure. Finally, heart acts as pressure source (sink) for venous return. (From Philip JH: A model for the physics and physiology of fluid administration, J Clin Monit 5:123, 1989.)

Fig. 55-3. PFR for five catheter sizes (16 × 2 inches, 18 × 2 inches, 20 × 1.25 inches, 20 × 1 inch, 24 × 1 inch). Pressures were measured at flows of 0, 100, 200, and 300 ml/hour, and resistance was computed as slope of least-squares regression line. Resistances are tabulated in Table 55-2. (From Philip JH: A model for the physics and physiology of fluid administration, J Clin Monit 5:123, 1989.)
the height of the free-air surface of the fluid above the reference level, less the height of the intervening air gap in the drip chamber.

IV tubing behaves as a low-resistance conduit relative to other system components:

$$R_{\text{tubing}} = 3 \pm 2 \text{ (SEM) RU}$$

A roller clamp behaves as a variable resistor. Once set, a roller clamp behaves the same ($\pm 5\%$) when connected to a flow source (IV pump) as with a pressure source (elevated bag). Typical roller clamp resistance during normal IV flow is 300 to 800 RU, depending on flow rate desired.

An IV catheter behaves as a resistor that offers more resistance than IV tubing. In the flow range of 0 to 300 ml/hour, the PFR is linear and resistance takes on a unique value for each catheter size.

<table>
<thead>
<tr>
<th>Gauge</th>
<th>Catheter plus tubing (R $\pm$ SEM)</th>
<th>Catheter alone (R $\pm$ SEM)</th>
</tr>
</thead>
<tbody>
<tr>
<td>16</td>
<td>6 $\pm$ 1</td>
<td>3 $\pm$ 2</td>
</tr>
<tr>
<td>18</td>
<td>10 $\pm$ 1</td>
<td>7 $\pm$ 2</td>
</tr>
<tr>
<td>20</td>
<td>17 $\pm$ 2</td>
<td>14 $\pm$ 3</td>
</tr>
<tr>
<td>22</td>
<td>34 $\pm$ 1</td>
<td>31 $\pm$ 2</td>
</tr>
<tr>
<td>24</td>
<td>66 $\pm$ 3</td>
<td>63 $\pm$ 4</td>
</tr>
<tr>
<td>None</td>
<td>3 $\pm$ 2</td>
<td>0 $\pm$ 2</td>
</tr>
</tbody>
</table>

*Resistance (R) in resistance units (RU) = 1 mm Hg/L/hour.

The PFR for systems composed of an IV catheter (Fenwall QuickCath) plus a tubing system (Travenol 2C0001) is plotted in Fig. 55-3. Measured total resistances ($\pm$ SEM) for no. 16, no. 18, no. 20, no. 22, and no. 24 catheters are 6 $\pm$ 1, 10 $\pm$ 0, 17 $\pm$ 2, 34 $\pm$ 1, and 66 $\pm$ 3 RU, respectively. When measured tubing resistance (3 $\pm$ 2 RU, previous equation) was subtracted from each, respective mean catheter resistances were 3, 7, 14, 31, and 63 RU (Table 55-2).

A normal patient vein behaves as a resistor, with resistance similar to that of a 20-gauge IV catheter. The distribution of patient resistance for the veins of 46 surgical patients is shown in Fig. 55-4. The population statistics for normal vein resistance were $R_{\text{vein}} = 22 \pm 20$ (SD) RU (range, 0 to 91 RU; median, 22 RU)

In addition to resistance, veins showed a second parameter that limited flow. In some patients, this was caused by external compression between the IV site and the heart (see later section on obstructions). In other patients, this pressure was the CVP itself. For the 46 veins studied, opening pressure was $P_o = 15 \pm 8$ (SD) mm Hg (range, 2 to 35 mm Hg).

The central venous circulation offers essentially no resistance to fluid flow. All resistance encountered distal to the catheter tip is attributed to the peripheral venous system. However, the central venous system does provide a back pressure equal to CVP. This pressure must be considered when assessing fluid flow through the system.

![Fig. 55-4. Distribution of hydraulic resistance measured in 46 veins and 12 infiltrations. Veins, open bars; infiltrations, closed bars. (From Philip JH: A model for the physics and physiology of fluid administration, J Clin Monit 5:123, 1989.)](image-url)
PATHOLOGIC SITUATIONS
Fluid Infusion into Tissues

The extravascular tissue near a vein behaves as a resistor that offers significantly more resistance than the vein. In the limited populations reported in the literature, tissue resistance is significantly greater than vein resistance. In addition, pressures measured at low flow are generally not elevated greatly above zero-flow pressures.

For tissues, resistance ($R_{tissue}$), estimated $P_o$, and initial pressure after infiltration ($P_{init}$) were $R_{tissue} = 1125 \pm 137$ (SD) RU, $P_o = 44 \pm 61$ mm Hg, and $P_{init} = 8 \pm 8$ mm Hg. Fig. 55-4 shows the distribution of tissue resistance superimposed on the distribution of vein resistance just described. $R_{vel}$ and $R_{tissue}$ were different significantly when tested with Student's t test ($p < 0.001$) and confirmed with Mann-Whitney U test ($Z = -5.31$). $P_{init}$ for tissues was not significantly different from $P_{o, vein}$. Volunteers remarked that infiltration at a low infusion rate (and thus low pressure) was not painful.

Obstructions to Fluid Flow: Starling Resistors

Many clinical situations produce vein collapse, causing the vein to behave as a Starling resistor. An inflated blood pressure (BP) cuff behaves as a Starling resistor, with $P_o$ close to the external pressure applied and little, if any, change in vein resistance. The PFR for a BP cuff on a typical patient is shown in Fig. 55-5. A venous tourniquet applied proximal to the IV site likewise behaves as a Starling resistor, raising $P_o$ to 25 to 45 mm Hg and not affecting $R$ significantly. Fig. 55-6 shows the PFR for a typical tourniquet-obstructed vein.

In vitro studies have shown that an elevated collapsible tube behaves as a Starling resistor, with $P_o$ approximately equal to the elevation of the collapsed segment. Also, an externally compressed collapsible tube behaves as a Starling resistor, with $P_o$ equal to external pressure.

Little information or data are available concerning IV sites that function erratically. It is known that a problem IV site can behave as a Starling resistor, with high obstructing pressure. In the one case reported, an IV catheter was in place but was not running well ("problem IV"). Although the IV bag was 80 cm above the patient’s heart and IV site, no fluid flow was detectable. A small fluid bolus injected via syringe into the IV tubing flowed easily, but spontaneous gravity-powered flow did not follow, even after catheter manipulation. When the IV bag was elevated 30 cm (12 inches), flow was observed to begin. After the IV site was determined to be functioning acceptably with bag elevated, the PFR was measured. Fig. 55-7 shows that resistance was 129 RU and $P_o$ was 80 mm Hg equals 109 cm $H_2O$. Analyzing the clinical situation and experimental data suggests that opening pressure for the Starling resistor was higher than the initial bag height, resulting in no flow initially. Elevating the bag slightly raised infusion pressure and resulted in significant flow.

A catheter manipulated against a vein wall also can behave as a Starling resistor. In the one case reported,
Fig. 55-6. PFR for typical patient vein obstructed by clinician-applied venous tourniquet. Note that applying tourniquet raises $P_o$ from 12 to 32 mm Hg. Resistance is relatively unaffected. (From Philip JH: A model for the physics and physiology of fluid administration, J Clin Monit 5:123, 1989.)

"normal vein" resistance was $R_{\text{vein}} = 30 \pm 9$ (SD) RU and $P_o = 7 \pm 1$ mm Hg. When a "positioned IV" was created by manipulating the catheter against the vein wall, $R_{\text{vein against wall}} = 27 \pm 0$ (SD) RU and $P_o = 36 \pm 0$ mm Hg.

Fig. 55-7. PFR for "problem IV" site in patient. (From Philip JH: A model for the physics and physiology of fluid administration, J Clin Monit 5:123, 1989.)

SYSTEM NONLINEARITY

When IV flow rate is low (less than 3000 ml/hour for most catheters and systems), the PFR is linear. However, when flow is increased significantly, the PFR loses its linear shape. Nonlinearity greatly affects flow prediction during rapid fluid infusion, as is required during volume resuscitation.

In 1983 Philip and Philip\textsuperscript{27} showed that the PFR for IV tubing systems and other fluid conduits is distinctly nonlinear and can be represented as:

$$P = (R_L \times F) + (R_T \times F^2)$$

Fig. 55-8 shows this. In the same study, by sequentially removing tubing sections, the authors showed that the $R_L$ and $R_T$ (multipliers of flow and flow squared) parameters are intrinsic properties of the tubing and do not result from flow perturbations at the entry or exit of the tubing system. They also showed that:

$$P = A \times F^n$$
does not fit the experimental data without systematic bias. Whether the observations represent concomitant laminar and turbulent flow in the same tubing section has not been established.

Finally, Philip and Philip\(^27\) showed that stopcocks and check valves are almost exclusively \(F^2\) devices, whereas 5 \(\mu\) filters (MP-5, Travenol) are exclusively linear (\(F\) only) devices. Later, the same authors showed that an \(F^3\) term is also required in analyzing the PFR for IV catheters.\(^{28}\)

To understand the impact of interposing or removing devices to enhance fluid flow, the system's nonlinearity cannot be ignored. The nonlinear PFR characteristics can be used to predict flow capability in IV infusion systems and assist clinical decisions on devices to be included or eliminated.\(^{29}\)

System nonlinearity can be ignored when flows are low enough. In 1988 Philip and Philip\(^{30}\) tried to identify situations in which nonlinearity was not significant. To do so, they defined \(F_E\) as the flow at which pressure drop was twice the linear pressure drop. At flows less than 0.1 \(F_E\), the system could be considered linear, and flow could easily be predicted from pressure by linear extrapolation. At flows above 1/10 \(F_E\), the effects of nonlinearity must be considered. Later, the order of device removal or replacement was assessed using increased flow capability as a criterion.\(^{29}\) The order was (1) fluid warmer removal, (2) 16-gauge to 14-gauge catheter change, (3) standard-bore to wide-bore tubing change, (4) 14-gauge to 12-gauge catheter change, (5) stopcock removal or change, (6) 12-gauge to 10-gauge catheter change, and finally (7) catheter elimination with the tubing directly connected to the patient's vein. Interestingly, removal of the stopcock was not required until the infusion system was composed of only wide-bore tubing and a 12-gauge catheter.

**CLINICAL FLUID INFUSION**

**Gravity Flow**

During routine anesthesia and surgery, fluid requirements in adults range from approximately 100 ml/hour to 100 ml/minute and more. Before anesthesia, an IV catheter is inserted in a vein and fastened securely, usually with adhesive tape. The size of the IV catheter is chosen to allow for the patient's expected fluid requirement. However, one should always recognize that rapid blood loss can occur during almost any surgical procedure and that the anesthesiologist must be prepared for such an occurrence.

When selecting catheter size, one must consider known information about the resistance of catheters, IV systems, and patient veins. The resistance of each of these flow impediments adds to the others to form the total resistance to flow. Thus total resistance to fluid flow \(R_{\text{total}}\) is the sum of the resistance of IV tubing, catheter, intervening devices, and the patient's vein.

The 22 RU median value for vein resistance and Table 55-2 (catheter resistances) together offer insight into the choice of catheter size. For a typical vein with resistance \(R_{\text{vein}} = 22\) RU, a 20-gauge catheter \((R = 17\) RU) approximately doubles resistance and halves flow. Changing from a 20-gauge to an 18-gauge catheter decreases total resistance from 39 RU to 32 RU, producing a resistance decrease of 18%, which causes a flow increase of 18%. Increasing catheter size
from 18 gauge to 16 gauge decreases resistance and increases flow by 13%.

The typical (22 RU) vein cannulated with a 17 RU, 20-gauge catheter provides a total resistance of 39 RU. If the pressure bag or bottle is 110 cm above the patient's heart, infusion pressure equals 110 cm H2O minus CVP (approximately 10 cm H2O), yielding a 100 cm H2O pressure head. Since 1 mm Hg = 1.34 cm H2O and 1 cm H2O = 76 mm Hg, pressure head = 76 mm Hg. Thus: F = P/R = 76 mm Hg/39 mm Hg/L/hour = 1.95 L/hour, or approximately 2000 ml/hour. For a 10 drop/ml drip chamber: Drop rate = 10 drops/ml x 2000 ml/hour x 1 hour/3600 seconds = 5.6 drops/second. This usually forms a steady stream. If a clinical IV infusion seems to flow much more slowly than this, the impediment imposed by the patient's venous system probably is greater than the typical 22 RU used in this example. Most likely a very small vein has been cannulated or the cannulated vein is not functioning normally. In conclusion, when a small IV catheter (20 gauge) produces low flow of IV fluid, the major flow impediment is probably not the IV catheter but rather the patient's vein.

With gravity as the driving force for fluid flow through an IV system, a major limitation in flow arises from the limited pressure head available. The IV bag or bottle can typically be placed no higher than 1 m above the patient's heart. This height limits the hydrostatic pressure that can be applied by gravity alone. However, since the pressure that reaches the patient's vein is often much lower than that applied by the pressure source because of resistive losses in the catheter and other components, higher pressures are usually safe. This is especially true when a small catheter size has been selected and a large vein is cannulated. In this situation, positive pressure can be applied, as described in the next section.

High-Flow Resuscitation

Rapid fluid resuscitation is often required during anesthesia and surgery. Since both pressure and resistance limit flow, both should be optimized to accomplish rapid fluid flow. Resistance is lowered initially by removing restrictions imposed by roller clamps. Ideally, only one roller clamp is present between bag and patient to reduce the likelihood that a second control will be left unadjusted. Resistance can be further reduced by removing interposing devices that may contribute to total resistance. When nonlinearity is considered, the order of device removal or change is (1) remove the blood warmer, (2) change catheter from 16 to 14 gauge, (3) remove check valve, (4) change catheter from 14 to 12 gauge, (5) replace regular tubing with wide-bore tubing, (6) change catheter from 12 to 10 gauge, (7) remove stopcocks, (8) remove the 10-gauge catheter, and (9) insert the sterile tubing end directly in the vein.

As an alternative to reducing resistance, pressure can be increased to increase flow. Pressure can be increased in either of two ways. Some IV sets contain an integral manual pump that allows the clinician to produce liquid flow by volume displacement. This labor-intensive approach allows bolus administration at any time, completely under the anesthesiologist's control. Also, this technique allows fluids contained in glass bottles (e.g., 5% albumin) to be infused with assistance.

An alternative approach increases infusion pressure with a pressure infuser, a pneumatic device that encircles the fluid-filled IV bag. Many competing devices are available. The maximum pressure suggested by manufacturers is usually 300 mm Hg, although much higher pressures are possible (JH Philip, unpublished data, 1990). Several different models of pressure infusers are available.

With the original pressure-infuser design, the clinician uses a manual bulb-inflator pump similar to that used with a sphygmomanometer cuff. Once the infuser is filled with air under pressure, infuser pressure drops as fluid is infused and leaves the bag, which is encircled by the infuser. Often a snap clamp is applied to the tubing between the bag and inflation bulb because of the misconception that infuser pressure falls as a result of leaks in the inflation bulb. Although inflation bulbs do leak, the ever-present leaks are typically small and are only significant when fluid is infused extremely slowly, as with an Intraflo (Abbott/Sorenson Co.) or other continuous low-flow device designed to infuse 3 ml/hour.

To maintain constant driving pressure for infusion, the pressure infuser can be attached to a constant pressure source. Again, the maximum recommended pressure is 300 mm Hg. When the infuser is used in this way, a constant pressure head is applied to the IV system-patient combination. This has the advantage of flow continuing at its initial rate, since additional gas enters the infuser to maintain a constant volume in the infuser-IV bag combination.

Whenever fluid is infused under pressure, increased vigilance is required to avoid infusion of air into the venous system. Despite the danger of air infusion, IV fluid manufacturers add potentially dangerous volumes of air to IV bags. Volumes as large as 70 ml may be present. Whenever fluid is administered under pressure, it is best to remove all air from the IV bag.

As the IV bag empties below 100 to 200 ml, most pressure infusers fail to apply their set pressure effectively to the bag, and infusion pressure drops. In an emergency situation, it might be best to discard the
residual 100 ml of fluid and change bags rather than try to infuse the last drop. Some infuser designs may be more effective, but such information has not been published.

Several pressure infusers can be attached to a single pressure source to facilitate rapid changing of crystalloid- and blood-containing bags during emergencies. One such device can provide 479 ml/min of lactated Ringer's solution or 318 ml/min of diluted packed RBC through standard IV tubing without a catheter. Several other devices have been described or are commercially available.

**MECHANICAL FLUID INFUSION SYSTEMS**

**IV Controllers**

Many devices are available to adjust or control the flow of fluid infused. IV controllers use gravity as a pressure source and manipulate the fluid path to adjust flow. Flow is usually monitored by an electronic drop sensor that causes the instrument to vary its slowing of the infusion.

Variable-resistance flow controllers vary the resistance of the infusion system by constricting or otherwise increasing the resistance of the tubing system.

Variable-duty-cycle controllers interrupt the application of pressure to the fully open IV tubing system and thereby control effective pressure applied from the fluid bag or bottle. The maximal effective pressure is limited by the bag height and also by the maximal duty cycle allowed by the infusion device.

Some IV infusion controllers are capable of monitoring the IV site for deterioration. Devices that quantify either the resistance imposed or the duty cycle applied have the capacity to perform monitoring. Some instruments sound an alarm when flow is too low, considering the instrument's purposeful limitation of flow.

Controllers' accuracy is limited by the flow-monitoring system's accuracy. Since many instruments use drop counting, variations in drop size described earlier play an important limiting role.

**IV Pumps**

Fluid pumps use positive displacement to provide regulated fluid infusion. Positive displacement can be provided by peristaltic fingers, a reciprocating syringe, or other mechanisms.

Because pumps use positive displacement, they may be capable of overcoming high resistance or back pressure to fluid infusion. This property can be advantageous in some situations. However, in the patient with fluid extravasation, some danger clearly exists.

Many pumps are incapable of detecting problems downstream from the pump unless these problems produce a significant pressure rise. Such pressure rise could result from total obstruction of the catheter or tubing. Situations such as fluid extravasation usually produce only a small rise in pressure because of resistance to infusion. Most pumps fail to detect such conditions. Complications of extravascular injection clearly become more problematic when the patient is unable to complain, as during general anesthesia.

**Other Fluids**

When fluids other than dilute crystalloid solutions are infused, the PFR may differ from that expected. The difference may be in the slope of the PFR line or in the shape of this relationship. When viscosity varies, the linear slope is expected to change. Differences in density could influence the nonlinear component.

Solutions containing dextrose (JH Philip, unpublished data, 1985) provide increased effective viscosity, as manifested by increased resistance with increased dextrose concentration. The specific relationship is:

\[ R_{D/w} = R_{\text{water}} (1 + C/1000) \]

where \( C \) is the dextrose concentration expressed in percent (%) and \( D/w \) represents dextrose with concentration \( C\% \) in water. (Actual slope of the regression line is 1.09 ± 0.05, with 95% confidence interval = 0.94 − 1.23).

Applying this relationship, the resistance of \( D_{25}W \) equals that of water \( \times 1 + 50^2/1000 = 1 + 2500/1000 = 1 + 2.5 = 3.5 \). Therefore, to achieve the same flow with \( D_{25}W \) as with water, pressure must be increased by a factor of 3.5. Alternatively, at the same pressure head, flow diminishes to \( 1/3.5 = 0.28 \) of the flow for dilute crystalloid.

From the resistance-concentration relationship, it can also be shown that relative viscosities of 1, 2, 3, and 4 are obtained with dextrose concentrations of 0%, 30%, 43%, and 52%. Similar analysis revealed that Intralipid has a relative viscosity of 1.36, equivalent to that of 17% dextrose in water. Osmolite (for gastric infusion) has a relative viscosity of 6.57.

**DRUG ADMINISTRATION**

The IV route is used extensively for drug administration during anesthesia. Drugs can be administered with several different infusion profiles. The most common drug infusion regimens are bolus injection and constant infusion. Occasionally, careful pharmaco kinetic control is used. The most common form uses an exponential decay in infusion rate to link bolus infusion to continuous infusion.
Delivery Devices
Many commercial devices are available for drug delivery, and several different technologies are available. Peristaltic pumps are usually reserved for nondrug fluid infusion because of the interface to an IV bag. However, many clinicians use a Buratrol or other device as a reservoir for the drug and then use a conventional fluid pump.

Syringe pumps appear better suited to drug infusion, since a small drug volume lends itself to containment in a syringe. Many competing devices are available.

Some drug infusion pumps are designed to facilitate drug administration on a milligrams/kilograms of weight basis. Others specifically facilitate use of individual drugs by using drug-specific labels. Still other drugs are capable of control by a computer that can infuse according to a prospective pharmacokinetic model or can vary drug infusion pharmacodynamically in response to monitored physiologic changes in the patient.

SUMMARY
IV therapy is an essential part of every anesthesia procedure. The IV infusion system connected to a patient can be analyzed according to simple physical and physiologic principles, and the resulting model can predict the IV infusion system behavior in many situations.

Devices are available to accommodate almost any need in fluid or drug administration. As new devices become available, the anesthesiologist must understand their operating principles and the physics that underlies them to recognize appropriate applications.

KEY POINTS
- The simplest IV system consists of an elevated, fluid-filled container, a length of flexible tubing, a catheter cannulating the patient's vein, and the patient's venous system.
- Flow through the IV system is determined by driving pressure and resistance of the catheter and tubing system.
- Driving pressure equals the height of the free-air surface of the fluid above the reference level, less the height of the intervening air gap in the drip chamber.
- IV tubing behaves as a low-resistance conduit relative to other system components: \( R_{\text{tubing}} = 3 \pm 2 \) (SEM) resistance units (RU). 1 RU = 1 mm Hg/L/hour.
- A roller clamp behaves as a variable resistor in that, once set, it behaves the same (± 5%) when connected to a flow source (IV pump) as to a pressure source (elevated bag). Typical roller clamp resistance during normal IV flow is 300 to 800 RU, depending on flow rate desired.
- An IV catheter behaves as a resistor that offers more resistance than does IV tubing, so that in the flow range of 0 to 300 ml/hour, the pressure-flow relationship (PFR) is linear and resistance takes on a unique value for each catheter size.
- A normal patient vein behaves like a resistor with resistance similar to that of a 20-gauge IV catheter. The central venous circulation offers essentially no resistance to fluid flow.
- The extravascular tissue near a vein behaves as a resistor that offers significantly more resistance than does the vein.
- Many clinical situations produce vein collapse, causing the vein to behave as a Starling resistor.
- When IV flow rate is low, the pressure-flow relationship is linear. When flow is increased significantly, the pressure-flow relationship becomes nonlinear and this significantly affects flow prediction during rapid fluid infusion, as is required during volume resuscitation.
- Great care must always be taken to avoid infusion of air into the venous system whenever fluid is infused under pressure; this is especially so since IV fluid manufacturers add volumes of air as large as 70 ml to IV bags, which preferably should be removed before use.
- Numerous mechanical systems are available to adjust or control the flow of the fluid infused. Simple controllers use gravity as the pressure source and a variable resistor to adjust the rate of infusion. The accuracy of controllers is limited by the accuracy of the flow monitoring mechanism, which often consists of a drop counting system that is limited by the variations in the size (and fluid volume) of the drops. Further, the pressure source varies because of patient movement and changes in the height of the bed.
- Fluid pumps use positive displacement to provide regulated fluid infusion. Positive displacement is
provided by peristaltic compression, by a reciprocating syringe, or by other mechanisms. Such systems can infuse fluid despite variations in downstream resistance, an effect that can be beneficial (e.g., by eliminating changes in fluid administration due to patient movement) or detrimental (e.g., subcutaneous extravasation of fluid), depending on the conditions. Many pump systems incorporate a pressure monitor that can detect complete obstruction of the outflow system but may not detect subcutaneous extravasation.

Several systems are available to facilitate intravenous drug administration. Many are now programmed to calculate the infusion rate, when data about patient weight, drug concentration, and desired dose (mg/kg) are provided. Although not foolproof, such devices should eliminate many of the risks of human error, especially in situations where the clinician has multiple tasks to perform simultaneously, as in complex anesthesia procedures.

**KEY REFERENCES**


**REFERENCES**

33. Philip JH, Philip BK: Hydrostatic central venous pressure measurement by IVAC


