Catheter Performance

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ABSTRACT

Venous catheters differ from peripheral arteriovenous (AV) access devices in many important ways. This discussion focuses on their performance as a conduit for blood flow between the patient and the dialyzer and on how catheter function is both limited and enhanced relative to the more common peripheral accesses. Catheter flow is limited by the high resistance inherent in the extended length of venous catheters relative to dialysis needles, but the high rate of flow in central veins also diminishes the opportunity for access recirculation. Cardiopulmonary recirculation is absent in patients with catheter access unless the patient also has a peripheral access. In the latter case, the same detrimental effect on urea clearance is seen regardless of which access device is used. Flow-dependent recirculation through circuits other than the peripheral AV access reduces the efficiency of dialysis (regardless of the type of access, catheter, or peripheral AV device used) across both catheters and peripheral AV devices. The inside diameter of the catheter plays a sensitive role in determining catheter resistance to flow. Slight increases in diameter under the same pressure head are associated with large increases in flow. Negative pressure at the catheter inflow port generated by the blood pump is magnified relative to peripheral devices, predisposing to partial collapse of the pump tubing segment and erroneous blood flow readings by the pump motor speed indicator. Setting a limit on prepump negative pressure can minimize this error. Future applications of dialysis may require lower pump speeds, which would allow more liberal use of catheter access if their potential for infection and clotting can be reduced.

The primary indication for use of venous hemodialysis catheters is for vascular access in patients with acute renal failure (ARF). While nephrologists try to discourage the growing use of catheters in patients with chronic or end-stage renal failure (ESRD), they welcome catheter access for patients with ARF because it allows rapid deployment of hemodialysis at reasonable blood flow rates (1–4). In recent years the increasing number of patients with acute need for dialysis, including those with ESRD who are poorly prepared for initiating their renal replacement programs, has enlarged the market for central vein dialysis catheters, promoting competition and improved catheter design. Among the many features sought by physicians who place and use these catheters is their performance, that is, their effectiveness as a conduit for blood flow between the patient and the dialyzer.

Catheter Resistance to Flow

Blood flow through the catheter is directly proportional to the pressure generated by the blood pump across the catheter and inversely proportional to resistance in the catheter itself:

$$Q_B \sim P/R \tag{1}$$

where Q_B is catheter flow, P is pressure, and R is resistance to flow. Modern blood pumps have a limited capacity to generate pressure, and high pressures can be damaging to red cells, so a major goal of catheter design has been to reduce the resistance to flow as much as possible.

Fortunately, the small diameter of catheters promotes laminar flow, which has lower intrinsic resistance compared with turbulent flow, where simultaneous lateral fluid motion impedes forward flow. Laminar flow can be predicted by the Poiseuille equation:

$$Q_B = k \cdot P \cdot D^4/(L \cdot V) \tag{2}$$

where k is a proportionality constant, D is the luminal diameter, L is catheter length, and V is blood viscosity. Comparison of Equation 2 with Equation 1 shows that resistance to flow is directly related to length and inversely proportional to the fourth power of the catheter diameter. The viscosity of blood at body temperature is relatively constant and is not easily modified, but the manufacturer can optimize both the *length* and the *diameter* of the catheter to reduce resistance. Doubling the length will double the resistance to flow, so if pump flow is limited by postpump pressure (due entirely to resistance in the catheter), the flow will be half that of the shorter catheter.

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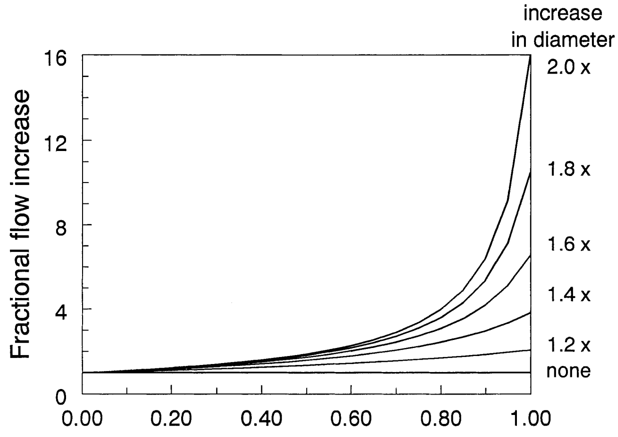


FIG. 1. When the catheter diameter is increased up to twofold, the effect on isobaric flow (vertical axis) depends on how much the catheter contributes to total resistance in the circuit (horizontal axis). If resistance within the catheter is the only resistance, doubling the catheter diameter may increase blood flow as much as 16-fold. If the catheter contributes only half of the resistance, flow can increase only twofold.

A more striking effect, however, is seen with changes in catheter diameter. Flow can be doubled when the diameter is increased by only 19%. A 50% increase in diameter increases flow fivefold, whereas doubling the diameter increases flow 16-fold if the catheter is the only source of resistance and all other parameters, including laminar flow, viscosity, and pressure, remain constant (Fig. 1). The relative resistance per unit of length is shown in Table 1 for catheters and needles that are currently used for hemodialysis vascular access. The smaller diameter of the needles used for peripheral arteriovenous (AV) access is offset by their shorter length, whereas the obligatory longer length of catheters requires a larger internal diameter to achieve adequate flow. Since catheters are placed in low-pressure veins where they are left for longer periods of time, the larger puncture hole is less of a problem than it is for peripheral AV grafts and fistulas that operate under higher pressures and must be repunctured for each dialysis.

TABLE 1. Effect of internal diameter on relative resistance to flow

	Diameter (mm)	Relative resistance per unit of length
PTFE graft	5.00 (4–7)	1
Standard cuffed catheter		
Adult	2.00	40
Pediatric	1.50	123
Tesio catheter	2.00	40
Quinton Marhukar	1.27	240
Thin-wall needles		
14 gauge	1.91	47
15 gauge	1.63	88
16 gauge	1.45	141
17 gauge	1.27	240

Importance of Prepump Pressure Monitoring

Most modern dialysis catheters are configured with double lumina to accommodate simultaneous inflow and outflow. This configuration facilitates placement with a single sterile needle puncture, but performance must be evaluated on both sides. On the outflow side, blood flow through the catheter is driven by the positive pressure generated by the blood pump, whereas on the inflow side, the driving force is the negative pressure generated by the same pump.

Venous Inflow

Because pressure in the central vein is close to atmospheric, the maximum negative pressure that can be generated on the venous inflow side (central vein to pump inflow) is −760 mmHg at sea level. However, the minimum pressure that most blood pumps are capable of generating is approximately −500 mmHg. Resistance to flow is almost entirely within the venous limb of the catheter, but a positive central venous pressure in the patient may help to maintain flow by preventing collapse of the vein around the catheter (5, 6). At the blood pump, this negative prepump pressure suddenly changes to a nearly equal but opposite (positive) pressure as the roller rotates off the pump trace (Fig. 2) (7).

Most roller pumps have only two rollers, so as the pump turns, one roller leaves the pump trace while the other enters it. Just before the first roller leaves the trace, the tubing segment between rollers is subject to negative (prepump) pressure, and just after it leaves the trace the same tubing segment is subject to positive (postpump) pressure. Typical pressures are −250 and +250 mmHg on the inflow and outflow sides, respectively, so the blood

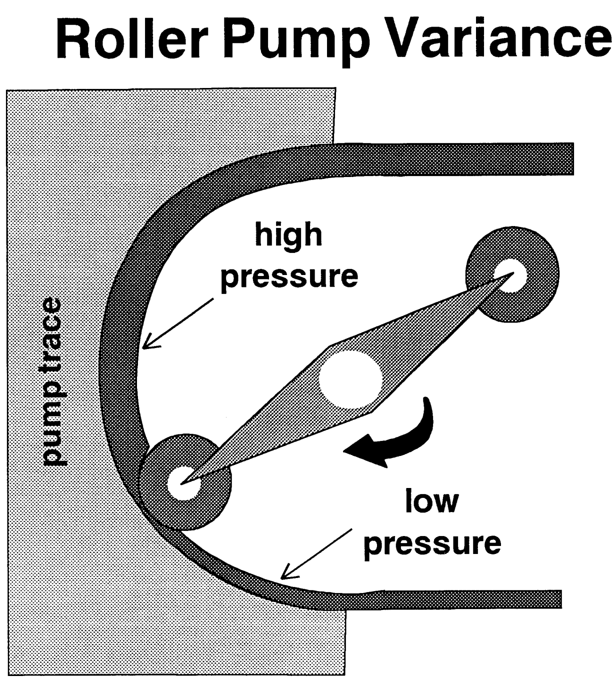


FIG. 2. Diagram of a hemodialysis roller pump. Adapted from T. A. Depner et al. (7), with permission.

in the tubing segment is typically subjected to a water hammer effect of 500 mmHg with each half turn of the pump. This pressure differential is greater for catheter access than for peripheral AV access because resistance to flow is higher through both limbs of the catheter. Flow in the pump segment and downstream may actually reverse directions briefly if the tubing segment between rollers is partially collapsed. The pump relies on the elasticity of the pump-tubing segment to expand and refill the segment on the inflow side while it is subject to significant negative pressure. Although the pump segment of the blood tubing is specially made from very resilient materials, refilling is less complete as prepump pressure falls, causing failure of the pump r.p.m. meter to accurately record blood flow. This effect is seen in Fig. 3, in which the relationship between true blood flow and flow calculated from the pump speed is shown at different levels of prepump pressure (7).

Despite manufacturers' efforts to employ tubing materials with minimal hysteresis, slight-to-moderate flattening of the tubing, which is warmed to body temperature during dialysis, has been implicated as a cause of reduced flow later on during a dialysis treatment (8). To prevent these errors caused by low venous pressure, the prepump pressure should be monitored and alarms set to stop the blood pump when venous pressure falls below −240 to −260 mmHg.

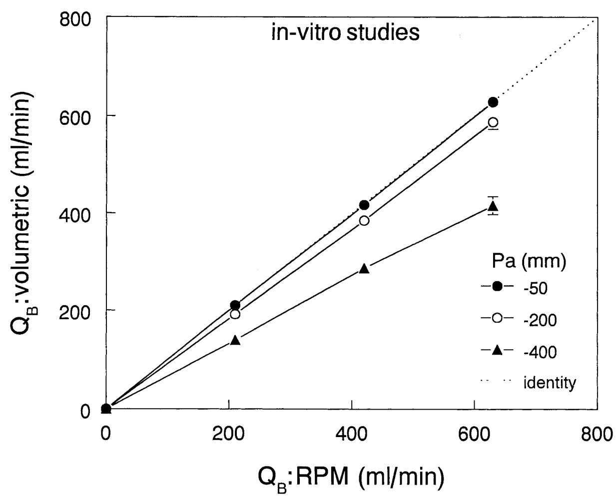


FIG. 3. True blood flow measured volumetrically in vitro (Q_B: volumetric) compared with flow calculated from the pump rotational speed (Q_B: r.p.m.) at different levels of prepump pressure (Pa). Reprinted from T. A. Depner et al. (7), with permission.

Arterial Outflow

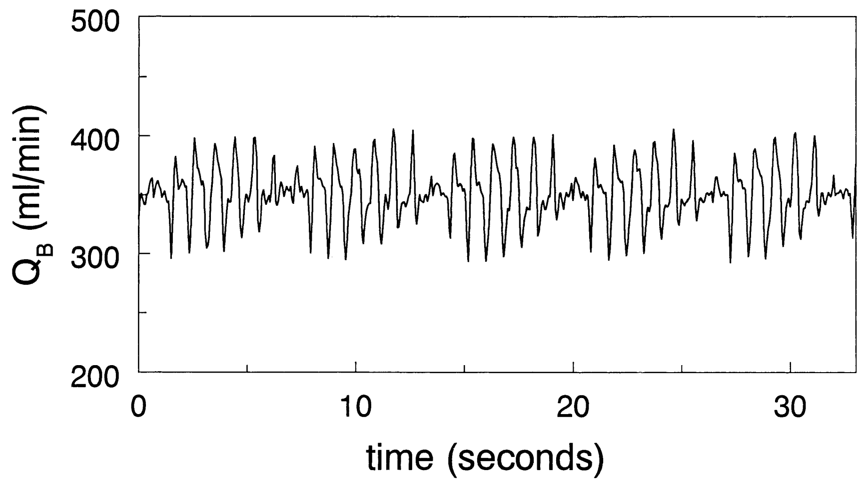
High pressures generated on the arterial side of the pump do not affect flow but can contribute to the above-water hammer effect that is commonly seen in the dialysis clinic as a rhythmic motion of the free tubing just beyond the roller pump. These pulsations, which are induced by oscillations in the blood tubing pressure downstream from the pump, are synchronous with the turn of the pump. At a blood flow of 350 ml/min, the pump turns at a speed of approximately 40 turns per minute, causing pressure oscillations at a frequency of approximately 80 per minute (Fig. 4), similar to the frequency of normal cardiac pulsations. Synchrony between the cardiac and roller pump pulses has been suggested as a possible cause of catheter recirculation (7, 9, 10).

The combined result of the above resistance and pump effects is to limit blood flow through dialysis catheters, often to rates considerably below that indicated by the pump speed meter. Rarely do flows exceed 400 ml/min, even in the largest catheters (11, 12).

Effect of Flow on Clearance

The catheter is a conduit from the patient's central veins to the dialyzer, where solute exchange occurs and toxins are removed. However, the fractional rate of removal (clearance) is not proportional to flow, as shown in Fig. 5. Within the range of clinically achievable blood flow rates (200–400 ml/min in catheters), sensitivity to flow is greater for high-efficiency dialyzers, i.e., those with larger surface areas and more permeable membranes (e.g., curve A in Fig. 5). The relationship between clearance and blood flow is also solute-specific; that is, it differs for different solutes moving across the same

FIG. 4. Pulsatile flow induced by the blood pump (see text). The amplitude modulation at a lower frequency is possibly induced by partial synchronization with the patient's own pulsatile cardiac output.



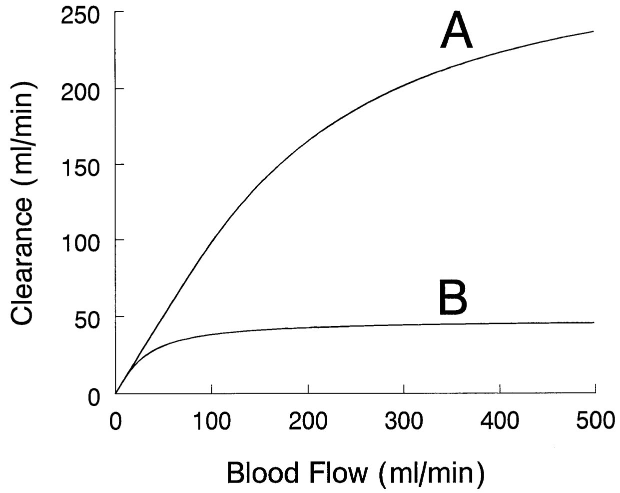


FIG. 5. Curves showing blood flow-dependent clearance (A) and blood flow-independent clearance (B).

membrane. In general, the clearance of smaller, more diffusible solutes depends more on blood flow than does the clearance of larger, less diffusible solutes.

These nonlinear effects thwart attempts to increase clearance by increasing blood or dialysate flow because clearance becomes a smaller fraction of flow at the higher flow rates. Conversely, for patients dependent on catheter access where blood flow is limited, the curvilinear relationship between flow and clearance is an advantage because the reduced flow is not accompanied by a proportionate reduction in clearance. For some solute/membrane combinations, little change in clearance occurs when blood flow is reduced (see Fig. 5, curve B). For these solutes, clearance becomes a higher fraction of the blood flow at lower blood flow rates; i.e., the efficiency of solute extraction increases. The reason for improved efficiency when flow is reduced is a matter of timing; time-dependent diffusion of solutes across dialyzer membranes is aided by slower blood flow rates. An example of a solute/membrane combination with so-called membrane-dependent clearance is shown in Fig. 5, curve B. For each stepwise decrease in blood flow, the clearance of the example solute decreases proportionately less than the clearance of a small solute like urea across a large permeable membrane (e.g., Fig. 5, curve A). Urea is an unusually diffusible solute with a clearance that is more flow-dependent than those of other solutes. The dialyzer extracts a higher fraction of the urea delivered to it, even at high blood flow rates. When dialyzer clearance is sensitive to blood flow, the fractional extraction of solute across the dialyzer is relatively independent of flow.

These relationships between blood flow and clearance assume that the dialysate flow rate is high. If dialysate flow is limiting, the relationship between blood flow and clearance of a small, easily dialyzed solute will look more like curve B in Fig. 5 than curve A. Conversely, when blood flow is limiting, increases in dialysate flow will have less of an effect on clearance. Since dialysate is relatively inexpensive and resistance to flow within the dialysate compartment is lower than resistance in the blood compartment, dialysate flows are ordinarily maintained at a level approximately twice that of blood flow.

Catheter Access Recirculation

When blood recirculates from the outflow port to the inflow port of a catheter placed in the femoral or iliac vein or across the ports of a malpositioned catheter in a vein other than the superior vena cava (SVC) or inferior vena cava (IVC), blood flow between the catheter ports has most likely reversed directions. This reversal of flow in the patient's vein is similar to recirculation in peripheral AV fistulas and is expected to occur whenever pumped flow through the dialyzer exceeds the flow in the vein. For catheters placed in the IVC or SVC, the causes of recirculation must be different, since flow in these large veins is unlikely to fall below the 300–400 ml/min maximum flow achievable through a catheter; i.e., reversal of bulk flow in the SVC or IVC is unlikely. Mechanisms that have been suggested to account for recirculation in larger veins are listed in Table 2. Since the velocity of flow in central veins is relatively slow compared with flow in the catheter, eddy currents are likely to occur, especially when the outflow port is partially obstructed by clot or the wall of the vein. Tricuspid insufficiency is also common, especially in volume-overloaded patients, so different causes may be operating in different patients.

Several older studies used an invalid method for measuring recirculation by comparing urea concentration in blood taken from the opposite arm to predialyzer and postdialyzer urea concentrations. This technique has been shown to routinely overestimate recirculation (7). Nearly all published studies of centrally placed catheters show average recirculation rates of less than 10%, suggesting that recirculation is indeed rare if the lines are not reversed (11, 13).

Quantifying Recirculation

Recirculation is usually expressed as a ratio or percentage of dialyzer blood flow. In other words, the flow of blood back through the inflow port directly from the outflow port is expressed as a fraction of the total pumped flow through the dialyzer. Since ultrafiltration within the dialyzer reduces the blood outflow rate, recirculation is conventionally expressed as a fraction of the inflow rate.

TABLE 2. Suggested mechanisms for catheter access recirculation in large high-flow veins

- Eddy currents caused by turbulent flow (perhaps induced by the catheter itself).
- Clotting at the tip of the catheter that may contribute to eddy currents or obstruct flow in the vein (17).
- Reversal of flow in the SVC during atrial systole (10).
  - Normal pulsatile flow
  - Tricuspid regurgitation
- Sheathing of the catheter with fibrin that might create a pathway from the outflow to inflow ports.
- Positive pressure ventilation (16).

Clinical Significance of Recirculation

Recirculation in catheters does not have the same significance as it does in peripheral AV access devices, where it strongly suggests stenosis in the vein. Whereas recirculation within a peripheral access necessitates angioplasty or surgical repair to salvage the access, the clinical significance of recirculation within a catheter is less clear. Catheter recirculation is increased by reversing the inflow and outflow blood lines and it reduces the effectiveness of dialysis (see below), but whether recirculation in catheters relates to poor cardiac output, valvular insufficiency, use of vasopressors, or other clinical states is not currently known, nor has it been investigated.

Long vs. Short Catheters

As discussed above, since longer catheters have greater resistance to flow, it would seem that shorter catheters are preferable. However, both upper-body catheters (e.g., internal jugular and subclavian) and lower-body catheters (e.g., femoral) may recirculate if they are too short because the volume flow of blood in smaller peripherally located veins is lower and can dip below the pumped flow rate (14). The flow in femoral veins at rest is relatively low because the major dependent organs requiring blood supply are leg muscles that are quiescent during dialysis. Femoral catheters should be at least 17 cm in length from the point of insertion in the groin to the catheter tip. Internal jugular and subclavian catheters should be long enough to extend to the right atrium, or at least to the junction between the right atrium and the SVC (15, 16).

Effect of Access Recirculation on Clearance

Access recirculation reduces the effective clearance of a solute by diluting the inflow concentration and thereby reducing the driving force for diffusion across the dialyzer membrane. Not all solutes are equally affected by a given percentage of access recirculation. Both a high dialyzer clearance (K_D), e.g., due to a high dialyzer mass transfer coefficient (K_0A), and a low dialyzer blood flow (Q_B) enhance the effect of recirculation on solute clearance because both of these factors enhance the solute extraction ratio (ER) across the dialyzer:

$$\mathrm{ER} = K_D/Q_B \tag{3}$$

Solutes with high extraction ratios (high K_D/Q_B) are affected more by recirculation than solutes with low extraction ratios (Fig. 6) because the solute concentration in the recirculated blood is proportionately more dilute. As shown by Equation 3 and in Fig. 6, the lower the dialyzer blood flow rate the greater the dilution effect, so recirculation occurring at low pumped flow rates has a greater effect on the clearance of all solutes, but especially those with high ERs, such as urea. This effect of blood flow has clinical significance because troubled catheters and AV fistulas limit dialyzer blood flow. When access recirculation appears in this setting, its effect on solute clearance is enhanced.

It should be noted, however, that although the relationship between extraction ratio (K_D/Q_B) and the

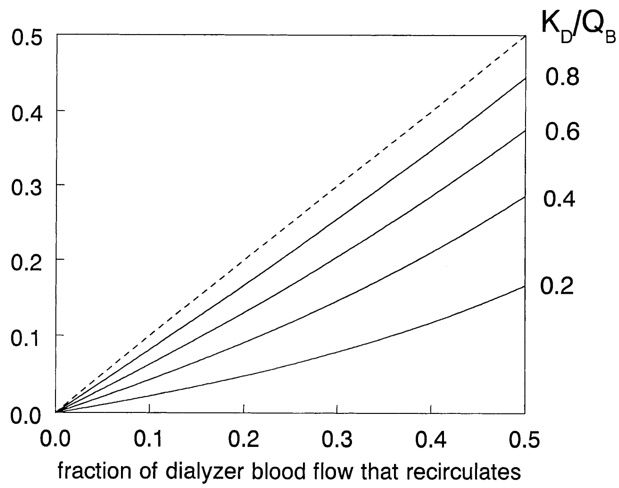


FIG. 6. The effect of access recirculation on clearance depends on the solute extraction ratio (K_D/Q_B) across the dialyzer. Reprinted from T. A. Depner (28), with permission.

reduction in clearance shown in Fig. 6 applies to all solutes, nearly all solutes have lower dialyzer clearances than urea, so the reduction in clearance will not be as great for these compounds as it is for urea; i.e., they will behave more like the lower curves in Fig. 6.

Reversing the Lines

Blood lines connected to the catheter inflow and outflow ports are often reversed by the dialysis technical staff to improve blood flow (Fig. 7). The precise reason for improvement in flow is not usually known in individual cases but is often thought to result from

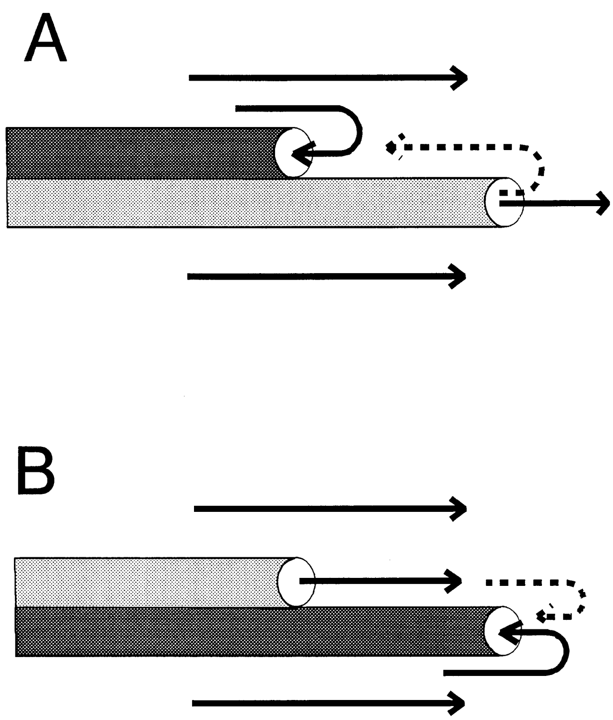


FIG. 7. Catheter blood flow in the normal configuration (A) and after reversal of the blood lines (B). The dashed lines show the pathways for access recirculation.

elimination of a ball/valve effect of flow in one direction. Another possible reason for improved flow is the potential for creating access recirculation. As noted above, this is less likely to occur in central vein catheters than in peripheral access devices because of the higher flow rates in the vena cava, but access recirculation has been demonstrated after reversing the lines. Twardowski et al. reported that recirculation averaged 7–14% when lines were reversed and that catheters with inflow failure had lower recirculation rates than well-functioning catheters (17). An extra 20–30 min of dialysis time was recommended to compensate for the adverse effect of reversing the catheters on solute kinetics.

Central Vein Compared with Peripheral Arteriovenous Access

Cardiopulmonary (CP) Recirculation

In patients with peripheral AV access devices, CP recirculation has a predictable influence on solute kinetics. CP recirculation is defined as flow of dialyzed blood back through the dialyzer via the central blood circuit (heart and lungs) without equilibration with blood in the rest of the body (7, 18). CP recirculation contributes to solute disequilibrium; that is, it causes solute concentration gradients to develop within the patient during dialysis. This form of *convective disequilibrium* differs from *diffusive disequilibrium* because it is caused entirely by differential blood flow rates throughout the body and is not influenced by solute diffusibility. For solutes that diffuse within the body less readily than urea, diffusive disequilibrium is enhanced but convective disequilibrium is equal to that of more diffusible solutes. This means that the fractional contribution of CP recirculation to total disequilibrium is reduced for less diffusible solutes. In concert with diffusive disequilibrium, CP recirculation accounts for much of the rebound in solute concentration following dialysis, accounting for approximately 30% of the total rebound of urea and less for other solutes.

The CP circuit can be visualized as one of several parallel vascular circuits that distribute blood to various parts of the body, each at different flow rates and with different transit times (7). The transit time through the CP circuit from output port to input port is short, often less than 10 seconds, while the average circulation time through the remainder of the body is about 1–2 min at rest. Blood in the CP circuit is able to flow again and again through the dialyzer, each time lowering solute concentrations, before blood flowing from remote compartments has an opportunity to return to the heart. The cause of this rapid flow is the peripheral AV access device, which provides a low-resistance pathway for blood pumped by the heart. The equilibrated clearance of urea is reduced approximately 10% because the low solute concentration in recirculated dialyzed blood dilutes the inflow solute concentration that serves as the source of energy for dialysis. The magnitude of this effect can be quantified as a function of dialyzer clearance (K_D) and cardiac output (CO):

$$\text{Fractional reduction in clearance} = K_D/(CO - Q_{AC} + K_D) \tag{4}$$

where Q_{AC} is blood flow in the AV access device. Equation 4 shows that states of low cardiac output and/or high flow in the peripheral AV access device will promote the reduction in clearance. High flow in the peripheral access device correlates with longevity of the device, but high flow also increases CP recirculation (19–22).

Fortunately, in patients with isolated central venous catheters there is no cardiopulmonary recirculation, so the efficiency of dialysis is increased somewhat above that of dialysis through peripheral access devices. Clearance is slightly enhanced and blood flow need not be as high in a patient with isolated catheter access to achieve the same clearance as in a comparable patient with a peripheral AV access. This partially offsets the lower clearance achievable through catheters due to their lower blood flow rates. However, the simultaneous presence of an AV fistula and other high-flow vascular circuits may cause recirculation of dialyzed blood in patients with catheter access.

Simultaneous Catheter Access and AV Fistula

Simultaneous presence of an AV fistula or graft in the same patient is common because patients often need temporary catheter access while a peripheral AV access matures. Compared with dialysis using the peripheral AV access device, the catheter simply changes the site where dialysis occurs within the cardiopulmonary circuit, so CP recirculation continues to reduce solute clearance. The low resistance of the AV fistula creates a high-flow pathway that carries dialyzed blood from the catheter through the fistula and back to the arterial inflow port where inflow solute concentration is reduced. This has the same effect on solute clearance as it does when dialysis takes place within the peripheral AV access device.

Other High-Flow Vascular Circuits

The second consideration is recirculation through high-flow vascular circuits other than the peripheral AV fistula (7). For example, flow through the liver or brain is relatively rapid and could have a diluting effect on the dialyzer inflow solute concentration and reduce effective clearance in a manner similar to CP recirculation. Some have suggested that catheters placed in the SVC may be less effective than those placed in the IVC because flow in the upper body tends to be more rapid than flow in the lower body (10). This theory also requires that the volume of the lower body (and the solute repository) be greater than that of the upper body served by the SVC. None of these theories have been tested.

Quantifying Effective Clearance in Patients with Catheter Access

The overall effect of diffusive and convective disequilibrium can be estimated as a function of single-pool clearance (spKt/V) and the rate of solute removal (K/V), which is a function of time on dialysis. Because of the reduced effect of convective disequilibrium, a separate equation is required for patients with central vein catheter access (23):

$$\text{Peripheral access: } eKt/V = spKt/V - 0.60(K/V) + 0.03 \quad (5)$$

$$\text{Catheter access: } eKt/V = spKt/V - 0.47(K/V) + 0.02 \quad (6)$$

where eKt/V is the effective dose of dialysis, K/V is (spKt/V)/t, and t is time on dialysis in hours. Equations 5 and 6 were derived empirically from population studies; each takes into account the reductions in clearance caused by diffusive and convective disequilibrium discussed above.

Recommendations

Catheter design should continue to focus on lowering flow resistance by increasing internal diameter rather than by shortening length. Length is required to ensure placement in the IVC or SVC as close to the right atrium as possible to avoid access recirculation. This is especially important for femoral placement. Additional studies are needed to examine the mechanisms responsible for recirculation in catheters placed in large, high-flow veins. Such studies might lead to further improvement in catheter design and placement. Prepump pressure in the inflow blood line should be monitored during dialysis to prevent excessively low pressures that collapse the pump segment and lead to erroneous measurements of blood flow. Access recirculation is best measured with an exogenous indicator such as normal saline, but such measurements are of less importance in patients with catheter access. Similarly, monitoring blood flow and/or venous pressure during dialysis in patients with catheter access is not as important as it is in patients with peripheral access devices, where it can extend the useful life of the access.

Anticipating future developments, the slower blood flow rates associated with the growing practice of extending time on dialysis and delivering it more frequently to patients with both acute and chronic renal failure may erase the inherent blood flow limitation of catheter access (24–27). Also, since uremic toxins are considered to be less diffusible than urea, less concern should be registered about access recirculation and cardiopulmonary recirculation. Both practices are predicted to have less effect on the clearance of less-diffusible solutes. Comparison of clearances in patients with catheters placed in the IVC versus the SVC might be helpful in determining potential advantages of catheter sites as they relate to solute clearance.

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