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CHAPTER 5

Continuous Kidney Replacement Simulator for Cumulative Balance and Measurement Redundancy

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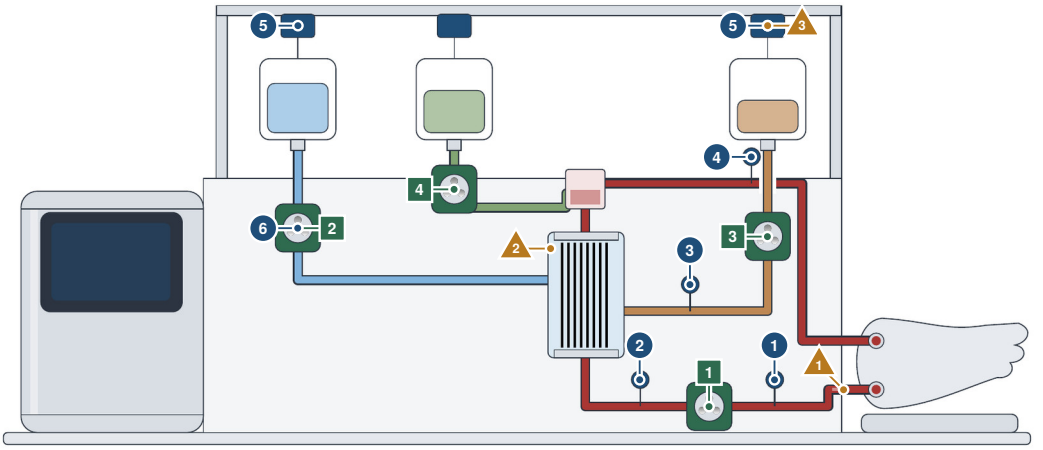
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CHAPTER 5

Continuous Kidney Replacement Simulator for Cumulative Balance and Measurement Redundancy



The controller moves 1 blood pump, draws blood through the filter 2 dialysate pump, feeds the filter jacket 3 effluent pump, takes the used fluid and the water 4 replacement pump, returns fluid to the blood

The controller reads 1 pressure in the access line 2 pressure before the filter 3 pressure in the effluent line 4 pressure in the return line 5 a scale holding each bag 6 the turns counted on every pump

What disturbs it 1 a narrowed catheter 2 a filter that clogs 3 a scale that drifts

The water the patient loses is what the effluent bag gains less what the two supply bags lose, and the scales and the pump counts measure the same flows twice. Illustration, not to scale.

IN PLAIN TERMS Weighing the Water a Machine Removes

The main idea. When the kidneys fail, a machine can clean the blood around the clock and take off extra water. It pumps liters of fluid in and out every hour. The water the patient loses is the small difference between the two.

The challenge. A tiny error in any pump grows large in that difference, so the machine weighs its fluid bags to check. If a scale slowly drifts, the machine trusts it and moves the patient off course while its own reading looks fine.

Why it matters. Over a day, those small errors decide how much water a very sick patient really loses.

THREE WAYS TO READ THIS CHAPTER

Explore	11 min	The first page, the fact box of Section 5.1, the history box of Section 5.2, the try-it boxes of Sections 5.4 and 5.6, the second fact box and the key points
Study	50 min	The whole chapter with its pause points and the twelve exercises
Practice	15 min	The first page, Section 5.6, Section 5.7, Section 5.8 and the open challenges

The control question The water a patient on continuous kidney replacement should lose is the small difference of liters an hour pumped in and out. What does a balance loop on a weighed running total remove, what does it pass to the patient while its own error stays small, and which second measurement shows the difference?

Learning objectives

- Compute how each pump's relative error is amplified in the net removal, and predict the removal error of a balance kept on pump counts.
- Derive the poles and the step response of a PI balance loop on an integrating running total.
- Explain why the balance loop follows a drifting scale, and size the smallest drift a plausibility check against the pump counts flags.
- Analyze how a shared, saturated effluent pump changes the sign of the dose loop's gain, and choose the structural remedy.
- Evaluate a circuit configuration over a day against removal, dose, pressure and ledger criteria.

Prerequisites Sections 1.6.2 (velocity form and back-calculation), 1.4.5 (redundancy), 1.4.6 (overrides on limits) and 1.6.4 (scenarios and the challenge).

The app https://apmonitor.com/apps/crrt-fluid-control/

See also Volume 4, Chapter 2 for a protective override that rests on its clamp and Volume 4, Chapter 12 for allocation of one limited resource. Volume 5, Chapter 13 for a running estimate that drifts until an independent fix arrives and Volume 5, Chapter 10 for a fast measurement corrected by a slower second one.

5.1 Introduction

When the kidneys fail in intensive care, continuous kidney replacement therapy removes waste solutes and water from the blood around the clock, often for days. The water matters as much as the solutes. In a multicenter observational study of critically ill adults with acute kidney injury, fluid overload of more than 10 % of baseline body weight was independently associated with mortality, although the study left open whether the overload caused the harm or marked a sicker patient [1]. The machine that delivers the therapy never weighs the patient. It knows the removal only through its own pumps and bags, and the removal it reports is a calculation on them.

DID YOU KNOW? Prescribed Is Not Achieved

In 500 critically ill adults treated over 3630 patient-days, with data from the health record and from the machines themselves, the therapy stood paused for a median 8.1 % of its time, and the achieved fluid balance missed the prescribed one by a median 17.4 mL per kilogram a day [2].

That calculation is a small difference between large numbers. In the first case of the simulator the blood pump counts 150 mL/min, or 9 L/h, through the filter, the dialysate and replacement pumps count 930 mL/h and 1000 mL/h into the circuit, the effluent pump counts 2030 mL/h out of it, and the patient is meant to lose 100 mL/h. An error in any of these flows reaches the removal whole, so each pump's relative error is amplified by the ratio of its flow to the removal, 20.3 for the effluent pump, 10.0 for the replacement pump and 9.3 for the dialysate pump. Roller pumps add a bias of their own, since they count revolutions and deliver less than the count when their inlet is under suction [3]. The tubing tolerances of the simulated circuit make the true removal 88.8 mL/h at reset while the pumps count 100 mL/h, and a balance kept on the pump counts leaves the patient 466 mL behind after a day.

Weighing the bags removes that bias and raises a second question. The simulator addresses the belief that a balance loop with a small error means a balanced patient. A loop that holds the weighed running total on its target rejects every pump error the scales can see, and it rejects a drifting scale just as thoroughly, by moving the patient until the reading is right. With the effluent scale drifting 60 g/h, 3 % of the effluent flow, the loop keeps its own measured error within 28.4 mL while the

true removal falls 304 mL behind the prescription in 6 h and 648 mL in 12 h. A measurement cannot check itself, and only an independent measurement of the same flow, here each pump's own count, reveals the drift, the redundancy of Section 1.4.5. A survey vehicle meets the same limit when its dead-reckoned position drifts until an acoustic fix arrives, as Volume 5, Chapter 13 shows.

Many plants share this shape. When a small net quantity is computed from large measured streams, the closure of the balance and the gross errors of individual meters decide what the result is worth, and adjusting redundant measurements to close material balances is the subject of data reconciliation in process plants [4]. A loss-in-weight feeder that holds its weighed flow but not the concentration of its product, in Volume 12, Chapter 12, is another case. A sensor fault is detected by comparing the sensor with a model or a second measurement of the same quantity, the basis of model-based fault detection [5]. Equipment limits that take a manipulated variable away from the loop that normally owns it are handled with override and selector control [6], and dialysis already uses feedback on a patient measurement to shape the removal, as when blood-volume tracking adjusts the weight-loss rate of intermittent hemodialysis [7]. However, a transfer function from pump speed to removal rate hides the running total, the subtraction and the instrument, and those three decide what a small error becomes by the end of a day.

This chapter uses a simulated continuous venovenous hemodiafiltration circuit, whose fluid balance is a running total assembled from four imperfect roller pumps and three scales, to show how the error each instrument makes reaches the patient while the loops see only what the instruments report. The therapy, gravimetric balancing, override control and plausibility checks are all in clinical or industrial use. The circuit model keeps its truth apart from its instruments, and its fluid ledger closes within 3×10^{-9} mL through flow, pauses and bag changes. Eight cases and a day-long challenge price a measurement, an override, a shared actuator and a pause in milliliters of removal error. The application is an educational equipment-control simulation. Its patient model, its filter coefficients and every clinical threshold it displays are illustrative and have not had clinical review, and nothing in it or in this chapter is clinical guidance.

Section 5.2 reviews how continuous therapy and its automatic balancing developed. Section 5.3 gives the circuit model, from the roller pumps to the scales and the ledger. Section 5.4 analyzes the balance as a small difference of large flows: how pump errors reach it, why the balance loop sees an integrator, why it follows a drifting scale and why the effluent pump is shared by three loops. Section 5.5 states the controller as implemented, and Section 5.6 reports the eight cases and the challenge. Section 5.7 describes the real scales, pumps and controllers, Section 5.8 states the limitations and Section 5.9 draws the results together.

5.2 Continuous Therapy and Automatic Balancing

Continuous therapy began as a way to remove water. In 1977 an arteriovenous hemofiltration circuit, perfused at about 100 mL/min by the patient's own arteriovenous pressure gradient with no blood pump, was reported to withdraw 200 to 600 mL/h from over-hydrated patients who were resistant to diuretics [8]. How the machines of that period came to balance the fluid themselves, first by the height of a bag and then by scales and pumps, is told in the history box of this section.

> ⧖ HISTORY **From the Height of a Bag to Scales (1977 to 1978)**
>
> The first continuous arteriovenous hemofiltration was given in 1977, in an intensive care unit in Göttingen, at a time when acute kidney failure was treated mostly with peritoneal dialysis or hemodialysis [9]. The removal was first controlled by hand. Hanging the filtrate bag higher or lower changed the suction of the fluid column below the filter, and the replacement fluid was also set by hand. Later systems balanced the fluid automatically, by gravity, with scales and electronic clamps, and peristaltic pumps soon replaced those simple mechanisms [9]. A clinical report of 1978 found automatic balancing accurate enough to replace the bed scales that conventional hemodialysis needed for patients confined to bed, and the constant weight loss seemed well tolerated by the patients [10].

Diffusion joined the continuous circuit a decade later. Continuous hemodialysis passed dialysate at 16.6 mL/min through a small flat-plate dialyzer in a pumpless circuit, and over blood flows from 70 to 190 mL/min the dialysate left in equilibrium with the blood, so the total clearance equaled the dialysate outflow of about 25 mL/min [11]. That equilibrium is why the clearance of a continuous therapy can be set by a fluid flow, and the diffusive term of the simulator reproduces it on a new filter. The limitations of the arteriovenous circuit led to pumped venovenous hemofiltration and hemodialysis, and collaboration with industry produced dedicated machines of greater accuracy [9]. Pumps brought an error of their own. In hemodialysis the blood flow measured by an ultrasonic sensor was 11.6 to 14.8 % below what the roller pump indicated at settings from 300 to 400 mL/min, and the shortfall followed the negative pressure in the arterial line [3].

The prescription became a rate per kilogram. A trial published in 2000 randomized 425 patients among ultrafiltration doses of 20, 35 and 45 mL/h per kilogram [12], and a multicenter trial in 2009 compared effluent flows of 40 and 25 mL/h per kilogram in postdilution continuous venovenous hemodiafiltration [13], the configuration simulated here. Both defined the dose by a fluid flow per kilogram. A machine delivers such a dose only while it runs and only as far as its filter still clears, which is why the simulator prescribes an effective dose as urea clearance per kilogram and scores its time average with the downtime included.

The patient side has dynamics of its own. Ultrafiltration takes water from the plasma, and the interstitium refills it at a rate described by Starling's hypothesis, with a filtration coefficient fitted to online blood-volume measurements during hemodialysis [14]. The circuit side has limits. Premature clotting shortens the life of the circuit, and the settings of the therapy, partial predilution and an individually controlled filtration fraction among them, are part of keeping it open [15]. A consensus on precision fluid management now frames fluid management during the therapy around three linked goals, the patency of the circuit, electrolyte and acid-base homeostasis and the regulation of the patient's fluid balance [16]. The simulator covers the first and the third and leaves electrolytes and acid-base out.

5.3 Circuit Model

The simulated circuit is a continuous venovenous hemodiafiltration with postdilution replacement on an illustrative 80 kg patient. Blood is drawn through the access lumen of a catheter by a roller pump, rises through the fibers of a hollow-fiber filter and returns through the return lumen, and the replacement fluid joins it in the venous chamber. Dialysate enters the filter jacket at the blood outlet end and flows counter-current to the blood, and the effluent pump draws the jacket out into a bag on a scale. Every flow, pressure, instrument and disturbance in the equations below is placed in Figure 5.1, and Table 5.1 classifies them.

5.3.1 Roller Pumps and Their Counts

Each roller pump delivers its stroke volume per revolution, reduced by its tubing's loading error, its wear and any suction at its inlet, as shown in Equation (5.1).

$$\begin{aligned} s_k &= V_{s,k}\,(1+e_{0,k})\max(0.5,\ 1-w_k N_k), \qquad \frac{\mathrm{d}N_k}{\mathrm{d}t} = n_k \\ Q_b &= s_b n_b (1+k_{in,b} P_{acc}), \qquad Q_d = s_d n_d, \qquad Q_r = s_r n_r \\ Q_e &= s_e n_e \max\bigl(0,\ 1+k_{in,f}\min(0, P_{eff})\bigr) \end{aligned} \tag{5.1}$$

where k is the blood (b), dialysate (d), replacement (r) or effluent (e) pump, $V_{s,k}$ the nominal stroke, $e_{0,k}$ the stroke error of the segment as loaded, w_k the stroke lost per revolution, N_k the revolutions turned, n_k the speed and k_{in} the under-filling per unit of inlet suction. The supply bags hang above their pumps, so only the blood and effluent pumps draw against suction. A supply pump delivers nothing once its bag is empty, and the effluent pump nothing once its bag reaches its physical limit.

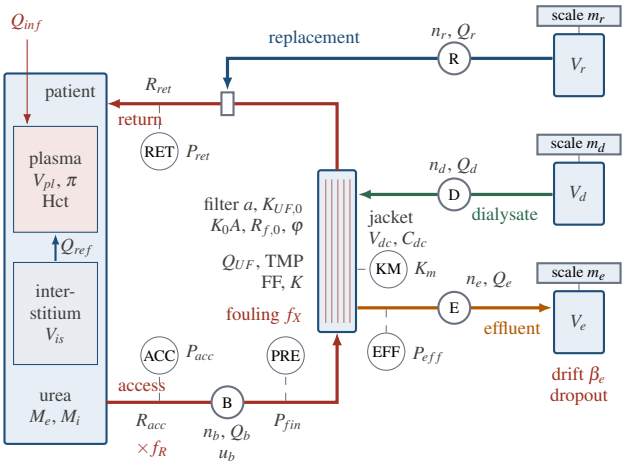


Figure 5.1 Continuous venovenous hemodiafiltration circuit as modeled. Blood leaves the patient through the access lumen of resistance R_{acc} (scaled by the restriction f_R), passes the access transducer ACC, the blood pump B and the pre-filter transducer PRE, rises through the fibers of a filter with open area a, and returns past the venous chamber and the return transducer RET. The dialysate pump D feeds the jacket counter-current to the blood, the effluent pump E draws the jacket (stored volume V_{dc}, pressure P_{eff} read by EFF) into the effluent bag, and the replacement pump R returns fluid to the venous chamber, or before the filter when predilution is chosen (not drawn). Each pump's encoder counts its revolutions N_k, each bag hangs on a scale reading m_k, and the idealized monitor KM reports the urea clearance. The patient is an illustrative plasma and interstitium with urea pools.

Table 5.1 Roles of the Circuit Variables at the Reference Configuration

Variable	Role	Reference	Range
$V_{net} - V_{Rx}$	CV, removal error	0 mL	±60 band
Q_b	CV, by pump count	150 mL/min	50–300
P_{acc}	constraint, override	−150 mmHg	−240 to −60
TMP	constraint, override	200 mmHg	100–300
FF	constraint, ceiling	25 %	10–35
D	CV, clearance per kg	25 mL/(kg h)	10–50
u_b, u_d, u_r, u_e	MVs, pump drives	%	0–100
$Q_{e,cmd}$	MV, balance	2030 mL/h	0–4500
Q_d, Q_r	MVs, dose, allocation	930, 1000 mL/h	0–4000
$Q_{net,Rx}$	prescription	100 mL/h	0–500
f_R, f_X	DV, restriction, fouling	1, 1	1–20, 0–20
β_d, β_r, β_e	DV, scale drift	0 g/h	±200
Q_{inf}	DV, other infusion	0 mL/h	0–200
n_k, N_k	states, pumps	4 each	—
V_{dc}, a	states, jacket, area	−0.21 mL, 1	—
V_d, V_r, V_e	states, bags	3.2, 4.1, 1.3 L	—
V_{pl}, V_{is}	states, patient water	3.2, 13 L	—
M_e, M_i	states, urea	30 mmol/L each	—
ledgers	states, integrals	9	—

Each drive follows its command through a lag and droops against the pressure it pumps into, as shown in Equation (5.2).

$$\tau_m \frac{\mathrm{d}n_k}{\mathrm{d}t} = n_{max,k} \frac{u_k}{100} \max\left(0.5,\ 1 - \delta \max(0, \Delta P_k)\right) - n_k \tag{5.2}$$

where the drive command u_k is in percent, the full speed is $n_{max,k} = Q_{max,k}/V_{s,k}$, δ is the droop and ΔP_k the pressure rise across the pump, $P_{fin} - P_{acc}$ for the blood pump, $P_{eff} - P_{head}$ for the dialysate

pump, $P_{ret} - P_{head}$ for the replacement pump and $-P_{eff}$ for the effluent pump, with P_{head} the head of the hanging bags. The machine's pump count is $V_{s,k}\hat{n}_k$, the nominal stroke times the speed its encoder measures over the last control interval, and it knows nothing of $e_{0,k}$, w_k or k_{in}. The loading errors of +0.4 % on the dialysate segment, −0.3 % on the replacement segment and −0.5 % on the effluent segment account for 10.9 mL/h of the 11.2 mL/h by which the true removal falls short of the count at reset, and the slight suction at the effluent pump for the rest.

5.3.2 Access, Filter and Return Pressures

The access lumen is a resistance in series with the blood pump, and Equation (5.3) solves the pump's inlet pressure and its delivery together.

$$P_{acc} = P_{cv} - f_R R_{acc} Q_b, \qquad Q_b = \frac{s_b n_b (1 + k_{in,b} P_{cv})}{1 + s_b n_b k_{in,b} f_R R_{acc}} \tag{5.3}$$

where P_{cv} is the central venous pressure and f_R the access restriction. Below the collapse pressure P_{floor} the line collapses, P_{acc} is held at P_{floor} and $Q_b = (P_{cv} - P_{floor})/(f_R R_{acc})$. At reset the blood pump counts 150 mL/min, delivers 144.7 mL/min and holds the access at −76.9 mmHg.

Filtration across the membrane follows a Starling law whose conductance falls as the filtered plasma concentrates [17], and the pressures that drive it follow from the flows, as shown in Equation (5.4).

$$\begin{aligned} Q_{UF} &= K_{UF,0}\, a \Big(1 - \frac{Q_{UF}}{\varphi Q_p}\Big)(\mathrm{TMP} - \pi) \\ \mathrm{TMP} &= \tfrac{1}{2}(P_{fin} + P_{ret}) - P_{eff}, \quad \pi = \pi_0 V_{pl,0}/V_{pl} \\ P_{ret} &= P_{cv} + R_{ret}(Q_b + Q_r - Q_{UF}) \\ P_{fin} &= P_{ret} + \tfrac{1}{2}(R_{f,0}/a)(2Q_{fin} - Q_{UF}) \\ Q_p &= Q_b(1 - \mathrm{Hct}) + Q_{r,pre}, \quad \mathrm{Hct} = V_{rbc}/(V_{rbc} + V_{pl}) \end{aligned} \tag{5.4}$$

In (5.4), a is the open share of the fibers, $K_{UF,0}$ and $R_{f,0}$ the ultrafiltration coefficient and blood-side resistance of the new filter, φ the filtration fraction at which the concentrated plasma stops filtering, π the oncotic pressure, $Q_{fin} = Q_b + Q_{r,pre}$ the blood entering the filter, $Q_{r,pre}$ the replacement flow when it enters before the filter and zero otherwise, and $V_{rbc} = V_{pl,0}\mathrm{Hct}_0/(1 - \mathrm{Hct}_0) = 1.37\,\mathrm{L}$ the fixed red-cell volume. Substituting the pressures gives $\mathrm{TMP} - \pi = A - R_p Q_{UF}$, with $A = P_{cv} + R_{ret}(Q_b + Q_r) + R_{f,0} Q_{fin}/(2a) - P_{eff} - \pi$ and $R_p = R_{ret} + R_{f,0}/(4a)$, so the first line of (5.4) is a quadratic in Q_{UF} whose smaller root is taken in closed form. For $A \le 0$ the membrane back-filters without throttling, $Q_{UF} = K_{UF,0} a A/(1 + K_{UF,0} a R_p)$, and with no plasma flow nothing filters. At reset TMP is 77.1 mmHg, the pre-filter and return pressures are 94.3 and 58.1 mmHg, 1086 mL/h crosses the membrane and the filtration fraction $\mathrm{FF} = Q_{UF}/Q_p$ is 17.9 %.

5.3.3 The Dialysate Compartment

The pumps impose the ultrafiltration, and the membrane settles at whatever pressure passes it. The jacket and the effluent line store fluid in a compliance, as shown in Equation (5.5).

$$\frac{\mathrm{d}V_{dc}}{\mathrm{d}t} = Q_d + Q_{UF} - Q_e, \qquad P_{eff} = \frac{V_{dc}}{C_{dc}} \tag{5.5}$$

where V_{dc} is the volume stored above the unpressurized jacket and C_{dc} the compliance. Linearizing (5.5) about the reset point gives the time constant $\tau_{dc} = C_{dc}/(\partial Q_{UF}/\partial(-P_{eff}) + \partial Q_e/\partial P_{eff})$. On the new filter the membrane passes 14.0 mL/h more for each millimeter of mercury drawn from the jacket, half the nominal coefficient because of the throttling and the return-pressure coupling in (5.4), the effluent pump's under-filling adds 0.4 mL/h per millimeter of mercury, and $\tau_{dc} = 62.4\,\mathrm{s}$. With the same pump flows at 15 % open area it is 326.6 s. Both are short against the hours over which the balance is judged.

5.3.4 Clearance and Fouling

Urea leaves the blood by diffusion into the dialysate and by convection with the ultrafiltrate, as shown in Equation (5.6).

$$\begin{aligned} K &= \min\{\lambda\,[Q_d(1-\mathrm{e}^{-K_0 A\,a/Q_d}) + S\,Q_{UF}^{+}],\ 0.9\,Q_{pl}\} \\ \lambda &= Q_{pl}/(Q_{pl}+Q_{r,pre}) \end{aligned} \tag{5.6}$$

where λ is the dilution by predilution fluid, one in postdilution, $Q_{pl} = Q_b(1-\mathrm{Hct})$ the plasma flow, K_0A the urea mass-transfer coefficient times area of the new filter, S the sieving coefficient and Q_{UF}^{+} the forward ultrafiltration. The effective dose is $D = K/W$ with W the weight. The diffusive term is the counter-current exchanger in the limit of a blood flow much larger than the dialysate flow, and it approaches Q_d when $K_0A\,a \gg Q_d$, the equilibrium seen in the first continuous hemodialysis [11]. At reset the clearance is 33.3 mL/min, of which 15.2 mL/min is diffusive and 18.1 mL/min convective, a dose of 25.0 mL/(kg h). The open area falls at a baseline fouling rate, faster at a high filtration fraction and while the blood stands still, as shown in Equation (5.7).

$$\begin{aligned} \frac{\mathrm{d}a}{\mathrm{d}t} = -a\big[&k_0 f_X + k_{FF}\max(0,\ \mathrm{FF}-0.2) \\ &+ k_{st}\max(0,\ 1-Q_b/Q_{st})\big] \end{aligned} \tag{5.7}$$

while $a > a_{min}$, where f_X is the fouling disturbance and $Q_{st} = 30\,\mathrm{mL/min}$. The coefficients K_{UF}, K_0A and $1/R_f$ of the filter all scale with a.

5.3.5 The Illustrative Patient

The patient is two fluid compartments and two urea pools, as shown in Equation (5.8), and it is illustrative.

$$\begin{aligned} \frac{\mathrm{d}V_{pl}}{\mathrm{d}t} &= Q_r - Q_{UF} + Q_{ref} + Q_{inf} \\ \frac{\mathrm{d}V_{is}}{\mathrm{d}t} &= -Q_{ref}, \qquad Q_{ref} = \frac{V_{is} - rV_{pl}}{\tau_{ref}(1+r)} \\ \frac{\mathrm{d}M_e}{\mathrm{d}t} &= G + K_c(C_i - C_e) - KC_e \\ \frac{\mathrm{d}M_i}{\mathrm{d}t} &= -K_c(C_i - C_e) \end{aligned} \tag{5.8}$$

where V_{pl} and V_{is} are the plasma and interstitial volumes, $r = V_{is,0}/V_{pl,0} = 4.06$ the ratio toward which the refill works, τ_{ref} the refill time constant, Q_{inf} any other infusion, M_e and M_i the urea in the extracellular and intracellular pools, $C_e = M_e/(V_{pl}+V_{is})$ and $C_i = M_i/V_{ic}$ their concentrations, G the urea generation and K_c the intercompartment clearance. The first line makes any departure from the starting ratio decay with time constant τ_{ref}, a first-order stand-in for the Starling refill that has been fitted to blood-volume data [14]. The patient exchanges fluid with the circuit directly, and the extracorporeal blood volume is not a state.

5.3.6 Bags, Scales and the Ledger

The bags integrate the pump flows, $\dot{V}_d = -Q_d$, $\dot{V}_r = -Q_r$ and $\dot{V}_e = Q_e$, and a bag change replaces a supply bag with a full one of volume V_{bag} or empties the effluent bag, recording what was taken down and what was hung. The scales read as shown in Equation (5.9),

$$m_k = \rho V_k + m_{tare} + b_k + \nu_k, \qquad \frac{\mathrm{d}b_k}{\mathrm{d}t} = \beta_k \tag{5.9}$$

rounded to 1 g, where $\rho = 1.00\,\mathrm{kg/L}$ for every fluid, m_{tare} is the bag and hook, b_k a bias that grows at the drift rate β_k and ν_k colored noise. A dropout event removes the reading altogether. The machine's net removal by the scales is the weight gained by the effluent bag less the weight lost by the two

Table 5.2 Model Parameters. Visible (vis.) Parameters Are Shown and Adjustable in the Model Panel, Hidden (hid.) Ones Are Not

Symbol	Meaning	Value	Class
W	weight, dose basis	80 kg	vis.
$V_{pl,0}$, $V_{is,0}$	plasma, interstitium	3.2, 13 L	vis.
Hct_0, τ_{ref}	hematocrit, refill	30 %, 40 min	vis.
P_{cv}, π_0	venous, oncotic	8, 25 mmHg	hid.
C_0, G	urea	30 mmol/L, 400 mmol/d	vis.
V_{ic}, K_c	urea pool, exchange	30 L, 600 mL/min	hid.
R_{acc}, R_{ret}	lumens	0.587, 0.35 mmHg min/mL	vis.
$R_{f,0}$	filter blood side, new	0.267 mmHg min/mL	vis.
P_{floor}	line collapse	−450 mmHg	hid.
$K_{UF,0}$	ultrafiltration, new	28 mL/(h mmHg)	vis.
K_0A	urea transfer, new	60 mL/min	vis.
φ, S, a_{min}	throttle, sieving, floor	0.70, 1.0, 0.03	hid.
k_0, k_{st}	fouling, stasis	2, 20 %/h	vis.
k_{FF}	per % of FF over 20	0.4 %/h	vis.
C_{dc}	jacket compliance	0.25 mL/mmHg	vis.
$V_{s,b}$, $V_{s,f}$	nominal strokes	7.5, 1.0 mL	vis.
$Q_{b,max}$	blood pump	450 mL/min	vis.
$Q_{d,max}$, $Q_{r,max}$	fluid pumps	4000, 4000 mL/h	vis.
$Q_{e,max}$	effluent pump	4500 mL/h	vis.
τ_m, δ	drive lag, droop	2 s, 3 % per 100 mmHg	vis.
$k_{in,b}$, $k_{in,f}$	under-filling	5, 2 % per 100 mmHg	vis.
w_b, w_f	wear	0.2, 0.15 % per 10^4 rev	vis.
$e_{0,d}$, $e_{0,r}$, $e_{0,e}$	loading	+0.4, −0.3, −0.5 %	vis.
$e_{0,b}$, P_{head}	blood error, bag head	+0.3 %, 20 mmHg	hid.
V_{bag}	supply and effluent bag	5 L	vis.
—	change levels	150, 4800 mL	vis.
t_{swap}	bag change	180 s	vis.
—	effluent limit, ramp	5.5 L, 30 s	hid.
m_{tare}, —	tare, resolution	250, 1 g	hid.
—	scale, pressure noise	2 g, 1.5 mmHg	vis.
—	transducer lag	1 s	vis.
—	monitor interval, noise	10 min, 3 %	vis.
W_c	check window	20 min	vis.
q_{abs}, q_{rel}	check tolerance	20 mL/h, 1 %	vis.
Hct_{ent}	entered hematocrit	30 %	vis.
—	TMP limit, stop	250, 350 mmHg	vis.
—	pre-filter warning, stop	300, 450 mmHg	vis.
—	access stop	−250 mmHg	vis.
ρ	every fluid	1.00 kg/L	fixed

supply bags, carried across each bag change by recording the final weight of the bag coming off and re-taring the new one. The true net removal obeys Equation (5.10).

$$V_{net} = \int_0^t (Q_e - Q_d - Q_r)\,\mathrm{d}t = -\Delta V_{pat} + V_{inf} - \Delta V_{circ} \tag{5.10}$$

where ΔV_{pat} is the change in $V_{pl} + V_{is}$, V_{inf} the other infusions and ΔV_{circ} the change in the circuit's stored volume, including what a discarded filter held. The prescribed trajectory $V_{Rx} = \int Q_{net,Rx}\,\mathrm{d}t$ is itself a state, and the removal error $V_{net} - V_{Rx}$ is what the scorecard judges. The ledger of patient, circuit, hanging bags, discarded bags and discarded filters, less the bags hung and the other infusions, stays constant within 3×10^{-9} mL in every run reported here.

5.3.7 Initialization and Numerics

Table 5.2 gives the parameters, and every clinical limit among them is illustrative. Every case starts with the treatment under way. Each pump turns at the speed that gives its prescribed flow by pump count, the jacket volume is found by bisection where $\dot{V}_{dc} = 0$, the filter is new, the bags hold 3.2, 4.1 and 1.3 L, and the patient sits at refill equilibrium with urea at C_0 in both pools. The pump strokes are

Table 5.3 Removal Error of Case 1 Over 8 h Against the Stroke Error of the Effluent Segment

Stroke error (%)	−1.0	−0.5[a]	0	+0.5	+1.0
Pump count, at 8 h (mL)	−184.6	−108.4	−33.1	+42.9	+117.6
Scales, worst (mL)	−9.7	−13.8	−10.2	−10.6	−9.5

[a] the loaded segment of the application

taken as already learned against the scales on earlier bags. Twenty-six states are integrated together by classical fourth-order Runge-Kutta at 0.5 s. They are the four speeds and four revolution counts, V_{dc}, a, the three bags, V_{pl}, V_{is}, M_e, M_i, five flow ledgers, V_{Rx} and the integrals of clearance, urea removed and urea generated. Flows and pressures are algebraic in the states, and bag swaps, filter changes and the hard limits of a full or dry bag are discrete events after each step. With the instruments noise-free, halving the step changes the removal error of case 1 at 4 h by 0.002 mL and its TMP by 0.0001 mmHg, and the TMP of case 3 at 6 h by 0.0002 mmHg, far inside the application's own check of 1 mL, 0.5 and 1 mmHg.

5.3.8 Assumptions

The patient is illustrative. It has two fluid compartments with a first-order refill and two urea pools with constant generation, and no blood pressure, tolerance of fluid removal, electrolytes, acid-base or drug clearance, so the model is silent wherever the patient's response to removal matters.

The membrane is lumped. Its filtration coefficient, urea transfer and blood-side conductance are proportional to one open-area fraction whose loss is prescribed. Real fouling is a protein layer and clot that also changes the sieving, and only urea is followed, with a sieving coefficient of 1. The concentration of the plasma along the fibers is a throttling factor rather than a distributed model, and the oncotic pressure follows the plasma protein alone.

The pumps are linear. Delivery is proportional to speed, wear to revolutions and under-filling to suction, where real tubing need not behave linearly at high suction. All fluids weigh 1.00 kg/L, so a scale reading in grams is a volume in milliliters. The clearance monitor reports the true clearance with noise, whereas real circuits more often infer clearance from blood and effluent samples. Anti-coagulation, the extracorporeal blood volume and the saline that primes a new filter are outside the ledger, and the blood is taken as returned before every filter change.

5.4 A Small Difference of Large Flows

Four properties of the balance follow from the model, and each shapes the controller.

5.4.1 How Pump Errors Reach the Removal

At the rates of case 1 the removal responds to a relative error ε_k in pump k as $\partial Q_{net}/\partial \varepsilon_k = \pm Q_k$, and relative to the prescription the amplification is $Q_k/Q_{net,Rx}$, 20.3 for the effluent pump, 10.0 for the replacement pump and 9.3 for the dialysate pump. A balance on the pump counts inherits every such error. Table 5.3 runs case 1 for 8 h with the effluent segment's stroke error set from −1 % to +1 %. On the pump count the error at 8 h moves 151.1 mL per percent, close to 1 % of the 15.6 L the effluent pump moves in that time, and it is −33.1 mL even with a perfect effluent segment, because the other segments, the wear and the suction still bias the count. On the scales the worst error of the same runs stays between 9.5 and 13.8 mL, whatever the segment does.

🕮 **Pause.** Try Exercise 5.2 before reading on.

5.4.2 The Balance Loop Sees an Integrator

The balance loop trims the effluent pump, and the weighed running total integrates the trim. Because the scales weigh the bags, the measured total answers the pumps within the 2 s drive lag, and the jacket lag of Section 5.3 acts on the patient and on TMP, not on the measurement. The control

Table 5.4 Scale Supervision Against a Constant Drift of the Effluent Scale Starting at 1 h, Case 5

Drift (g/h)	Flagged after onset	Error at 12 h (mL)
20	never	−223.3
30	never	−322.7
35	never	−379.3
40	25.0 min	−44.7
60	16.0 min	−33.2
80	12.0 min	−46.4
60, supervision off	—	−647.8

model is therefore an integrator, $V_{net,m}(s) = \Delta(s)/s$ in the trim Δ, and a PI controller with gain K_c and integral time τ_I gives the closed-loop characteristic polynomial $s^2 + K_c s + K_c/\tau_I$. The defaults, $K_c = 2/\text{h}$ and $\tau_I = 1\,\text{h}$, place the poles at $-1 \pm \text{j}$ per hour, a damping ratio of 0.707 and a natural frequency of 1.41 rad/h. A step ΔQ in an unmeasured rate then leaves the error $\Delta Q\, e^{-t} \sin t$, with t in hours, which peaks at $0.322\,\Delta Q$ after 0.785 h.

To test that model, the 150 mL/h step of case 2 was applied at 0.25 h with the feedforward off, the trim bound raised to 1000 mL/h and the instruments noise-free. The linear prediction is a peak of 48.4 mL at 0.785 h after the step. The engine's error is already 2.2 mL behind when the step comes, because the loop is still absorbing the 11.2 mL/h shortfall it started with, and it peaks 53.4 mL behind 0.79 h after the step. Taken as the stepped run less an otherwise identical unstepped run, the step response peaks at 49.6 mL after 0.80 h. The remaining 1.2 mL comes from the effluent segment's under-filling, which grows as the added effluent flow draws the jacket down from −1.5 mmHg to about −15 mmHg at the peak, so the pump delivers less than each added milliliter of command. With the fluid segments' under-filling set to zero the step response peaks at 48.4 mL after 0.786 h, as predicted, while setting the loading errors to zero changes nothing. The shipped case 2 makes the step at 1 h with noise and a trim bound of 300 mL/h, and its true error peaks 65.8 mL behind at 1.782 h, as the pumps restart after the effluent bag change that began at 1.722 h. At that instant the controller's own error is 53.4 mL, the scales read the running total 2.8 mL low, and the working target trails the trajectory by the 15.2 mL of backlog that the pause has just accrued. Raising the bound to 1000 mL/h leaves that peak unchanged. The time scales are well separated, since the drives settle in seconds, the jacket in 62 to 327 s and the balance in about an hour, so each layer can be tuned as if the faster ones were instantaneous.

5.4.3 A Loop That Holds a Drifting Measurement

The integrator in the plant and the integral in the controller make the balance a type-2 loop, which follows a ramp without steady error. A scale bias that grows at a constant rate is such a ramp in the measurement, and the loop follows it exactly by moving the true removal instead. In case 5 the effluent scale drifts 60 g/h from 1 h, the loop keeps its measured error within 28.4 mL for the whole day, and the patient is 37.4 mL behind at 2 h, 304.3 mL at 6 h, 647.8 mL at 12 h and 1358.9 mL at 24 h. By 6 h the effluent scale has counted 292.3 mL of its 300 g bias into the running total. Of the other 7.7 g, 6.0 g accrued while its bag was off the scale during the changes at 1.72 and 4.20 h, and the rest is reading noise. No tuning of the balance loop removes the ramp-following error, since any stable PI loop on this integrator follows the ramp, and a slower loop only follows it later.

🕮 **Pause.** Try Exercise 5.1 before reading on.

The plausibility check of Section 5.5 compares each scale's flow over 20 min with its pump's calibrated count and flags a disagreement above 20 mL/h plus 1 % of the flow at two checks in a row, about 40 mL/h for the effluent scale. Table 5.4 shows what that tolerance lets through. A drift of 40 g/h or more is flagged within 25 min and the error at 12 h stays within 46.4 mL, while 20, 30 and 35 g/h are never flagged and leave the patient 223.3, 322.7 and 379.3 mL behind. Below the tolerance

the drift is also taught to the pump count. At each clean effluent bag change the calibrated stroke is re-learned from the scale, by at most 1 % per bag, so at 35 g/h it climbs from 0.995 to 1.012 by 4.23 h and the reference follows the fault. Near the tolerance the outcome depends on the noise. At one of four other seeds 30 g/h is flagged 51 min after the onset, at the check after the first effluent change, whose pause shrinks the flow part of the tolerance and whose final weighing and re-tare each add one noisy reading to the window, and the substitute count then carries the part of the drift already learned, so the error still reaches 181.5 mL by 12 h.

> **TRY IT IN THE APP** **5 · Scale Drift**
>
> Load scenario 5 and press Run, then open the Balance view. From 1 h the effluent scale drifts. Watch the first trend: the true removal error leaves the ±60 mL band while the error by the scales, dashed, stays inside it. By 6 h the alarm badge reads a removal error of −304 mL.

5.4.4 One Pump for Every Flow

The effluent pump carries the dialysate, the replacement and the net removal at once, so its speed limit is shared by three loops. Below the limit more dialysate is matched by more effluent, and the steady gain of the dose loop is the diffusive slope $\partial K/\partial Q_d = 1 - \mathrm{e}^{-x}(1+x)$, with $x = K_0 A\, a/Q_d$, which is 0.897 at the reset of case 1. At the limit the allocation of Section 5.5 protects the net removal, so each added milliliter of dialysate displaces a milliliter of replacement and, with $S = 1$, a milliliter of convective clearance, and the gain becomes roughly the diffusive slope less one. At the moment the effluent pump reaches full speed in case 8, at 3.330 h, the steady gain computed from the model is 0.427 while the effluent follows the dialysate and -0.493 once the replacement gives way, and at 7 h, with the dialysate at its limit, it is -0.820. A dose loop whose process gain changes sign becomes a positive feedback that drives the dialysate up while the dose falls. The remedy is structural, a cap on the dialysate that keeps effluent headroom, and no tuning of the dose loop supplies that headroom, since a slower dose loop only postpones the fall. Allocation of one limited resource among several demands is the subject of Volume 4, Chapter 12.

Pause. Try Exercise 5.3 before reading on.

5.5 Controller as Implemented

The engine of Section 1.6.1 runs the controller once per second, every second physics step, before the integration and the discrete events of that step. The dose loop acts only when the clearance monitor reports, every 10 min. The historian logs one row every 20 s, 4320 rows of 147 columns in a day, among them the true running total beside the totals by the scales and by the pump counts, the pump commands, speeds and true deliveries, the pressures, the three scale readings with the effluent scale's bias, and the supervisor's state. The application runs from 1 to 3600 times real time, 300 by default, at which a four-hour case takes 48 s and the day-long challenge 4.8 min.

5.5.1 Instruments

No loop reads a bag volume, a flow or the patient. Four pressure transducers read the access, pre-filter, return and effluent pressures through a 1 s lag with colored noise of 1.5 mmHg correlated over 2 s, and TMP is formed from them as in (5.4) and filtered through 20 s for the override. Four encoders count revolutions exactly. The scales follow (5.9) with 2 g of noise correlated over 3 s and a 250 g tare. An idealized clearance monitor reports the true clearance with 3 % noise every 10 min while the circuit runs, never within 5 min of a restart. Two detectors report an effluent bag at its limit and air from an empty supply bag.

5.5.2 The PI Algorithm

Every loop is an instance of the framework's velocity-form PID of Section 1.6.2, Equations (1.12) to (1.15), with the derivative time at zero and no rate limit. The chapter writes the action as $\mathscr{A}$, the

Table 5.5 The Loops as Configured. Every Loop Is PI, With No Derivative Action and No Rate Limit

Loop	PV → MV (limits)	K_c	τ_I	T_t, $\mathscr{A}$
Blood speed	$\hat{n}_b \rightarrow u_b$ (0–100 %)	2.0 %/rpm	3 s	τ_I, +
Fluid speeds	$\hat{n}_k \rightarrow u_k$ (0–100 %)	1.5 %/rpm	3 s	τ_I, +
Access	$P_{acc,m} \rightarrow$ speed cap	0.10 rpm/mmHg	10 s	10 s, −
TMP	$\mathrm{TMP}_f \rightarrow$ UF ceiling	20 mL/h/mmHg	600 s	60 s, +
Balance	$V_{net,m} \rightarrow \Delta$ (±300)	2.0/h	1 h	τ_I, +
Dose	$D_m \rightarrow Q_d$ (0–$Q_{d,lim}$)	60 kg	1 h	τ_I, +

a of that section: $\mathscr{A} = +1$ for a positive process gain, which is reverse acting, and $\mathscr{A} = -1$ for a negative one, which is direct acting. Only the access override is direct acting, since a faster blood pump draws the access pressure lower. The tracking time T_t of the back-calculation is set per loop in Table 5.5. The control period Δt_c is 1 s for every loop but the dose loop, which uses 600 s at each monitor sample. In MANUAL the four drive stations belong to the operator, each speed loop sets its integral so that its output equals its station, the balance and dose loops track what the stations amount to and the two overrides wait at their ceilings, so the return to AUTO is bumpless. Table 5.5 lists the loops.

5.5.3 Speed Loops and the Access Override

Each pump holds its speed on its encoder. A fluid pump's setpoint is its flow command divided by the nominal stroke, and the blood pump's is the lower of $Q_{b,SP}/V_{s,b}$ and the speed ceiling from the access override, a PI on the measured access pressure against its limit of −150 mmHg. The speed loops remove the droop of (5.2), and they cannot remove the under-filling of (5.1), since the pump turns at the right speed and moves less. An override that is not limiting waits just above the demand it may have to take over. The access ceiling sits at 110 % of the speed the flow setpoint asks for, held there by back-calculation with a 10 s tracking time, so it binds within seconds of the pressure crossing its limit instead of unwinding from full speed. Section 1.4.6 places such overrides on the limits of a living plant, and Volume 4, Chapter 2 treats a protective override whose output rests on its clamp until a limit nears.

5.5.4 Balance, Feedforward and Backlog

The balance loop compares the measured running total $V_{net,m}$, by the scales or by the pump counts at the operator's choice, with the working target $V_w = V_{Rx} - B$, and its output Δ trims the effluent command, as shown in Equation (5.11).

$$Q_{e,cmd} = Q_d + Q_r + Q_{ff} + \Delta, \qquad Q_{ff} = Q_{net,Rx} + \dot{B}_c \tag{5.11}$$

with Q_{ff} the feedforward from the prescription, zero when the feedforward is off, and $\dot{B}_c$ the catch-up rate. While the fluid pumps are paused, the backlog B grows at $Q_{net,Rx}$, so the working target stops and the loop does not wind up against a stopped pump. After the restart, with catch-up on, B is repaid at up to $r_c = 60\,\mathrm{mL/h}$ and the repayment is fed forward. The trim is bounded to ±300 mL/h and to the room that the effluent pump and the ultrafiltration ceiling leave. When the feedforward is turned on or off, the integral absorbs the change in the feedforward term at the same instant, and the effluent command stays where it was.

5.5.5 Filter Protection and the Allocation of the Effluent Pump

Ultrafiltration is capped by the lower of two ceilings, the output U_T of a PI on the filtered TMP against its setpoint of 200 mmHg, and the filtration-fraction ceiling $\mathrm{FF}_{max}\hat{Q}_p$ with $\hat{Q}_p = \hat{Q}_b(1 - \mathrm{Hct}_{ent}) + Q_{r,pre}$ built from the blood pump's count and an entered hematocrit. When not limiting, the TMP override waits 300 mL/h above the ultrafiltration it may have to cap, with a 60 s tracking time, and it

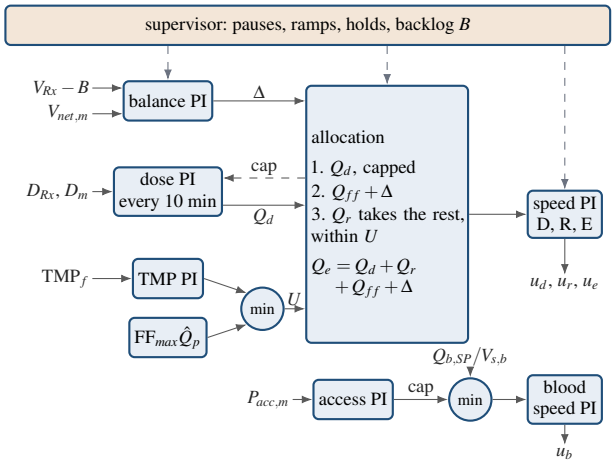


Figure 5.2 Loops above the pump drives. The balance trim Δ, the dialysate from the dose loop and the ultrafiltration ceiling U meet in a fixed allocation of the effluent pump, in which the replacement receives only what the dialysate, the net removal and the ceiling leave, and the headroom option caps the dialysate. The measured running total comes from the scales or the pump counts and, with scale supervision on, a flagged or silent scale is replaced by its calibrated count.

is frozen for 3 min after a restart while the jacket refills. The effluent pump's capacity is then allocated in a fixed order, as shown in Figure 5.2 and Equation (5.12).

$$\begin{aligned} Q_r &= \mathrm{clamp}\big(Q_r^*,\ 0,\ Q_{r,max}\big), \quad U = \min(U_T,\ \mathrm{FF}_{max}\hat{Q}_p) \\ Q_r^* &= \min\{Q_{r,Rx},\ U - Q_{ff} - \Delta^+,\ R_e - \Delta\} \\ R_e &= Q_{e,max} - Q_d - Q_{ff} \end{aligned} \tag{5.12}$$

where $\Delta^+ = \max(0, \Delta)$. The dialysate is set first, the net removal and its trim come next, and the replacement receives what remains up to its prescription, so an override or a full effluent pump takes convection before it takes removal. With the headroom option on, the dialysate is capped at $0.9\,Q_{e,max} - Q_r - Q_{ff}$ using the replacement of the previous scan, which keeps 10 % of the effluent rating for the trim, the wear and the suction.

5.5.6 Dose Tracking and Honest Shortfalls

The dose loop moves the dialysate to hold the measured dose $D_m = K_m/W$ at its prescription, updating once per monitor sample and holding its output between samples and through pauses. At the operating point of case 1 each 1 mL/h of dialysate adds 0.897 mL/h of clearance, 0.0112 mL/(kg h) for the 80 kg patient, so the default gain gives a proportional loop gain of 0.67 per sample. The machine reports a shortfall instead of forcing it. It declares the removal unreachable after 5 min with the ultrafiltration room exhausted or the effluent pump and trim at their limits, and the dose unreachable after 30 min with the dialysate at its limit and the monitor below 97 % of the prescription.

5.5.7 Scale Supervision

With scale supervision on, each scale's running total is compared once a minute with its pump's calibrated count, as shown in Equation (5.13).

$$\delta q = \frac{(S_b - S_a) - (C_b - C_a)}{W_c}, \quad \tau_q = q_{abs} + q_{rel}\,\frac{|C_b - C_a|}{W_c} \tag{5.13}$$

where S and C are one-minute means of the scale's running total and of the calibrated count at the two ends of the window $W_c = 20$ min, with $q_{abs} = 20$ mL/h and $q_{rel} = 1$ %. Two consecutive checks with

$|\delta q| > \tau_q$ flag the scale, and the balance then uses the calibrated count for that bag, rebuilt from the start of the window. A silent scale is replaced by the count at once, and a recalibration re-bases every scale on the total in use. The calibrated count is the nominal stroke times a factor learned from the scale at each clean bag change of at least an hour, limited to 1 % per bag, times the suction correction $1 + k_{in,f} \min(0, P_{eff,m})$ for the effluent pump, so it follows the wear that the raw count misses. With supervision off, a silent scale holds its last reading.

5.5.8 Supervisor

A supervisor sequences the pumps through five states, RUN, STOPPING, CHANGING, ALARM HOLD and RESTART. A supply bag at 150 mL or an effluent bag at 4800 mL, as the machine estimates them, starts a bag change, and with the change-together option any other bag due within 45 min joins the same pause. The fluid pumps stop, the final weights are recorded, the bags are swapped after 180 s and re-tared, and the pumps ramp back over 30 s, about 3.7 min in all, while the blood keeps circulating. A measured access pressure below −250 mmHg for 2 s, or a pump-count blood flow under 30 % of its setpoint for 20 s, stops every pump until the operator restarts. A measured TMP above 350 mmHg for 10 s stops the fluid pumps, a pre-filter pressure above 450 mmHg for 5 s stops the circuit as clotted, and a filter change stops it for 20 min. While the state is not RUN the upper loops hold their outputs, the TMP override freezes and the missed removal accrues as backlog.

5.5.9 Simulation, Control and Internal Models

Three models are kept apart here. The simulated plant is the model of Section 5.3. The control model of the balance is the integrator of Section 5.4, whose gain is slightly below one because part of each added milliliter of effluent is lost to suction. The control model of the TMP override is the jacket lag with a steady gain from ultrafiltration to TMP that rises from 0.065 to 0.413 mmHg per 1 mL/h as the open area falls from 1 to 0.15, while its time constant rises from 62.4 to 326.6 s, so the override meets a plant that stiffens and slows as the filter fouls. The controller's internal models are nominal by design. The feedforward assumes every pump delivers its nominal stroke, the filtration-fraction ceiling trusts the pump-count blood flow and the entered hematocrit, the headroom trusts the nominal effluent rating, and the calibrated count trusts the learned stroke and the nominal suction coefficient. None of them knows the patient.

5.6 Cases and Measured Outcomes

Table 5.6 gives what the eight cases produced, and the subsections below give their scripted events. Each case begins from a reference configuration, with the balance on the scales, scale supervision, feedforward, catch-up at 60 mL/h, filter protection, dose tracking with effluent headroom and the access override on, and a removal of 100 mL/h, 930 mL/h of dialysate, 1000 mL/h of replacement and a dose of 25 mL/(kg h), apart from the changes named. Every value below comes from a headless run of the simulation engine at its default seed, with any setting changed before the run, and each is a true value rather than an instrument reading unless stated. Four other seeds move the noise-dependent results only modestly, the worst error of case 1 over 4 h between 7.9 and 10.9 mL and the flag of the case 5 drift between 14 and 16 min. Errors are quoted as the largest deviation of the true running total from its trajectory, which, unlike an integrated error, stops growing when a transient ends.

Each case is a comparison of two runs: a run as shipped, a Reset, which restarts from $t = 0$ and keeps the learner's settings, one changed setting and a second run. The challenge is judged on the case as given, and changing a model parameter or a disturbance voids the attempt. The briefings, completion rules and the challenge follow Section 1.6.4.

5.6.1 The Balanced Circuit and a New Target

On the scales, case 1 holds the running total within 7.9 mL of its trajectory over 4 h, through two bag changes that pause the fluid pumps for 7.4 min, and within 22.3 mL over a day of 18 changes. The day's worst comes at 14.41 h, as the third of three bag changes within 20 min ends with 11.4 mL of

Table 5.6 Measured Outcomes of the Eight Cases

Case	Measured outcome
1, scales	worst 7.9 mL in 4 h, 22.3 mL in 24 h
1, pump count	48.6 mL behind at 4 h, 466.3 mL at 24 h
2, feedback only	65.8 mL behind at 1.78 h, 62.2 mL ahead at 5.01 h
2, feedforward	worst 18.2 mL, 46 min after the change
3, protection on	TMP at most 207.5 mmHg, Q_r 797 at 6, 503 at 8 h
3, dose	25.0 to 19.6 at 6 and 13.7 mL/(kg h) at 7.99 h
3, protection off	TMP 250 mmHg at 6.05 h, 350.9 mmHg at 7.70 h, stop
4, override on	−150 mmHg, count 111.6 mL/min, true 103.5 mL/min
4, filtration fraction	true 25.4 %, estimated 24.0 %, peak 26.3 %
4, override off	−198.7 mmHg, count 150 mL/min, true 135.5 mL/min
4, kinked line	stop at 2.00 h, 14.9 min, area 93.3 to 88.2 %
5, supervision off	measured within 28.4 mL, true −304.3 mL at 6 h
5, supervision on	flagged after 16.0 min, −12.4 mL at 6 h
6, catch-up off	10 pauses, 37.2 min, −88.1 mL at 8 h
6, catch-up on	worst 18.5 mL, together 5 pauses in 18.6 min
7, hold last reading	measured −651 mL, true +79.8 mL
7, substitute count	worst 6.8 mL
8, no headroom	full speed at 3.33 h, 21.4 mL/(kg h) over hours 5 to 8
8, headroom	27.8 mL/(kg h) over hours 5 to 8, shortfall at 3.17 h

backlog to repay, while the noisy final weighings and re-tares of the earlier changes have left the scale total about 12 mL above the true one. At four other seeds it is 12.2 to 19.9 mL, and noise-free 10.8 mL. The trim averages 12.3 mL/h over the first four hours, close to the 11.2 mL/h shortfall at reset, and it grows with the wear to 34.5 mL/h over the last four hours of the day (12.9 and 29.8 mL/h noise-free). On the pump count the patient is 48.6 mL behind at 4 h and 466.3 mL behind at 24 h. In the first four hours the open area falls to 92.3 % at the baseline fouling rate, TMP rises to 83.6 mmHg and urea falls from 30.0 to 26.6 mmol/L.

Case 2 raises the removal to 250 mL/h at 1 h and lowers it to 50 mL/h at 4 h. On feedback alone the patient falls 65.8 mL behind at 1.78 h, the trim reaches its 300 mL/h bound at 1.86 h and stays at or near it until about 4 h, and when the target drops the integral is still carrying the old rate, so the patient runs 62.2 mL ahead at 5.01 h and 31.4 mL ahead at 6 h. With the bound raised to 1000 mL/h that overshoot is 63.1 mL, close to the linear peak of 64.5 mL that the step down of 200 mL/h alone gives, so the bound does not cause it. Fed forward, the prescription leaves the loop only the pump errors, and the worst error is 18.2 mL, 46 min after the first change. In that run the filtration fraction climbs to 22.3 % at 250 mL/h, the plasma falls to 2.89 L from 3.20 L at reset and the hematocrit rises from 30 to 32 %, because the interstitium refills the plasma more slowly than the circuit takes it.

5.6.2 Fouling and the TMP Override

Case 3 fouls the filter twelve times faster than the baseline. The open area falls to 61.9, 38.3 and 23.7 % at 2, 4 and 6 h, and TMP rises from 77.1 to 111.4 and 163.3 mmHg at 2 and 4 h, since TMP less the oncotic pressure grows about inversely with the open area. TMP reaches the override setpoint at 4.99 h, and from 5.16 h the override cuts the replacement command, to 797 mL/h by 6 h and 503 mL/h by 8 h, with TMP at most 207.5 mmHg. Dose tracking meanwhile raises the dialysate from 930 to 1630 and 2729 mL/h, and the dose falls anyway, from 25.0 to 19.6 at 6 h and 13.7 mL/(kg h) just before the bag change at 7.99 h, because diffusion through a clogged membrane saturates at $K_0 A\, a$, 8.8 mL/min at 8 h. At 8.2 h tracking is worth 1.0 mL/(kg h), 13.2 against 12.2 without it. An effluent-based dose would read 41.7 mL/(kg h) at 7.99 h, since the effluent carries 3339 mL/h, while the clearance delivers 13.7. The pre-filter pressure passes the 300 mmHg filter-spent warning at 7.81 h, the filter is changed at 8.25 h, the 20 min stop takes the removal from 13.6 mL ahead to 20.2 mL behind, and TMP is back to 84.0 mmHg at 8.7 h. With the protection off TMP passes 250 mmHg at 6.05 h and reaches 350.9 mmHg at 7.70 h, where the fluid pumps stop until the filter change, and the

removal is 103.5 mL behind when they restart at 8.59 h. With the protection on, the removal error stays within 13.9 mL until the filter change, because the replacement gives way first.

5.6.3 Access Restriction

Case 4 raises the access resistance 2.6-fold at 0.5 h. The override holds the measured access pressure at −150 mmHg by slowing the blood pump to a mean of 111.6 mL/min by count between 1.0 and 1.7 h, while 103.5 mL/min actually passes, since each revolution under that suction moves 7.3 % less. The filtration-fraction ceiling uses the count, so the machine's estimate averages 24.0 % against a true 25.4 %, and the true fraction peaks at 26.3 %. The ceiling is only as good as the blood-flow estimate. Without the override the pump keeps its 150 mL/min count, delivers 135.5 mL/min and holds the access at −198.7 mmHg. At 2 h the line kinks to 20 times its resistance, the access pressure falls to the −450 mmHg collapse pressure and the pumps stop at 2.00 h. The blood stands still for 14.9 min, the open area falls from 93.3 to 88.2 % and the removal falls 28.7 mL behind. The circuit is back in RUN at 2.26 h, catch-up repays the 25.8 mL backlog over the next 25.8 min, and without catch-up the patient would still be 37.4 mL behind at 4 h.

5.6.4 Drift and Dropout

With scale supervision on, the drift of case 5 is flagged at 1.27 h, 16.0 min after it began and with 16 g of bias, the balance is rebuilt on the pump count from the start of the window, and the patient is 12.4 mL behind at 6 h. The count cannot follow the wear between recalibrations, and the error reaches 36.8 mL by 12 h. The case is shown in Figure 5.3 as shipped, with the supervision off.

Case 7 silences the effluent scale for 20 min at 1 h. With supervision off the last reading is held, the measured removal falls by the inflow, and the loop sees the patient 651 mL behind by 1.33 h and drives the trim to its 300 mL/h bound from 1.07 h. The patient is in fact being over-dried, 79.8 mL beyond the prescription when the reading returns and 10.6 mL at 2.5 h. With the bound at 1000 mL/h the same dropout over-removes 209.0 mL, so the harm scales with the authority given to the outer loop. With supervision on the count stands in for the missing scale and the error never exceeds 6.8 mL.

> **TRY IT IN THE APP** **7 · Sensor Dropout**
>
> Load scenario 7 and press Run. At 1 h the effluent scale stops reporting for 20 min. With Scale supervision off the loop drives the effluent trim to its 300 mL/h bound, and the patient is 79.8 mL beyond the prescription when the reading returns. Press Reset, set Scale supervision to On and run again: the error never exceeds 6.8 mL.

5.6.5 Bag Changes

Case 6 prescribes 1800 mL/h of dialysate, 1200 mL/h of replacement and a 150 mL/h removal, so the effluent pump draws about 3150 mL/h and the effluent bag fills about every 1.6 h. Without catch-up, ten changes pause the fluid pumps for 37.2 min in 8 h and leave the patient 88.1 mL behind. With catch-up the backlog is repaid and the worst error is 18.5 mL. Changing every bag due within 45 min in the same pause halves the pauses to five and the paused time to 18.6 min, and raises the delivered dose from 33.13 to 34.33 mL/(kg h). The ledger closes within 3.1×10^{-10} mL through every change.

5.6.6 The Effluent Pump at Its Limit

Case 8 treats a 120 kg patient at 250 mL/min of blood, 1500 mL/h of dialysate, 2000 mL/h of replacement and a 150 mL/h removal, 3650 mL/h of effluent by count and a dose of 29.0 mL/(kg h) against a prescribed 30, and raises the prescribed dose to 40 mL/(kg h) at 1 h with the headroom off. The dose loop raises the dialysate until the effluent drive reaches full speed at 3.33 h, and from 3.42 h the effluent command sits at its 4500 mL/h limit. The replacement then gives way milliliter for milliliter, its command falling to 368 mL/h by 8 h, the dialysate reaches its 4000 mL/h limit at 6.44 h, and the

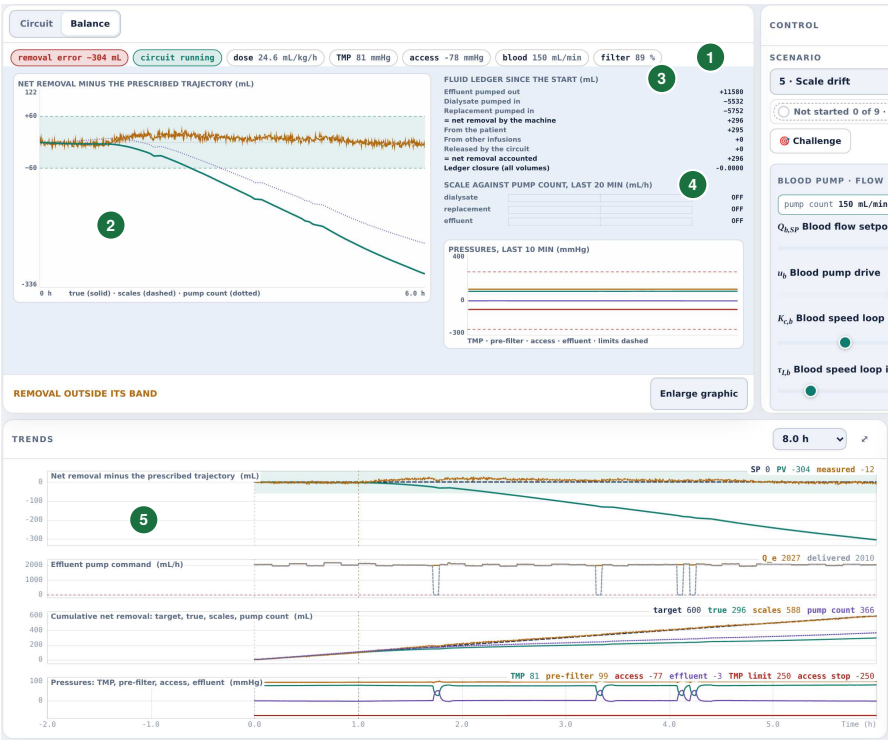


Figure 5.3 The application in case 5, scale drift, at $t = 6.00$ h with scale supervision off as shipped, five hours after the effluent scale began to drift at 60 g/h, on the Balance view, cropped to the plant view and the trends. Callouts: (1) the status badges, (2) the plot of the removal error in the scene, (3) the fluid ledger, (4) the scale checks and (5) the first trend. The plot in the scene and the first trend show the true removal error leaving the ±60 mL band while the error by the scales, dashed on the trend, stays inside it, and the alarm badge reads a removal error of −304 mL. The fluid ledger beside the plot balances 11 580 mL of effluent against 5532 mL of dialysate and 5752 mL of replacement, a net removal of 296 mL with 295 mL of it from the patient, and closes to 0.0000 mL, while the three scale checks read OFF. The trends over an 8 h window show the true error at −304 against a measured error of −12, the effluent command at 2027 mL/h (2010 mL/h delivered) with four dips for bag changes, and the running totals parting, target 600 mL, true 296 mL, scales 588 mL and pump count 366 mL. TMP is 81 mmHg on a filter 89 % open.

dose delivered over hours 5 to 8 is 21.4 mL/(kg h), below the 29.0 of the start. The worst balance error, 32.5 mL ahead at 7.11 h, builds while the dialysate drive runs at 100 % against a pressure rise of 53.8 mmHg and turns at the speed for 3936 mL/h of its 4000 mL/h command, since the allocation gives the replacement the room the command leaves. With the headroom on, the dialysate is capped near 1900 mL/h, the dose over hours 5 to 8 is 27.8 mL/(kg h), the machine reports at 3.17 h that the prescribed 40 is out of reach, and the worst balance error is 20.3 mL.

5.6.7 A Day on the Circuit

The challenge runs 24 h with the removal rising from 100 to 150 mL/h at 8 h, fouling at 3.2 times the baseline, a partly blocked catheter from 5 to 7 h, a kinked line at 12 h, an effluent-scale drift of 60 g/h from 15 h, a routine recalibration at 17 h and 40 mL/h of other infusions. It is met with the true running total within ±60 mL of its trajectory at every instant, at least 85 % of the prescribed dose delivered, TMP at most 250 mmHg, a filtration fraction at most 25 % and the ledger closed within 1 mL. Table 5.7 compares the configurations, and Figure 5.4 draws three of its columns as bars. In its

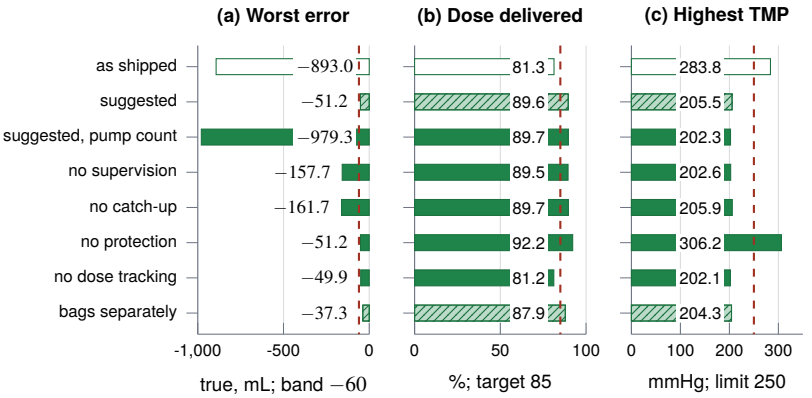


Figure 5.4 The eight configurations of Table 5.7 over the day of the challenge: (a) the worst deviation of the true running total against the ±60 mL band, (b) the dose delivered against the 85 % target and (c) the highest TMP against 250 mmHg, limits dashed. Rows after the second change one element of the suggested configuration. White: the shipped configuration. Solid: the other rows that fail. Hatched: the two rows that pass. Three of the failing rows fail on the band alone, the run without protection on TMP and the run without dose tracking on the dose.

shipped form the circuit balances on its pump counts with 700 mL/h of dialysate and no catch-up, filter protection or dose tracking, and it ends 893.0 mL short of its 3200 mL target, with TMP past 250 mmHg from 21.92 h and 81.3 % of the dose. The suggested configuration adds the scales with supervision, catch-up, protection, dose tracking and bags changed together. Its worst error is 51.2 mL behind at 12.26 h, as the circuit restarts after the access stop, 8.8 mL inside the band. It ends 24.1 mL short of the target, flags the drift at 15.27 h, holds TMP to 205.5 mmHg by cutting the replacement command to 617 mL/h by the end of the day while the dialysate rises to 3070 mL/h, pauses 14 times, 13 of them for bag changes and one for the access stop, for 63.7 min, keeps the filtration fraction at most 22.5 % and delivers 89.6 % of the dose. The application gives this one run whether the changes are made before Run or carried through a Reset. At four other seeds the suggested configuration still meets the target, with worst errors from 31.1 to 59.3 mL, the largest 0.7 mL inside the band, so its margin depends on the noise. Removing any single element fails except changing the bags together. Without the filter protection the dose is higher, 92.2 %, because the replacement is never cut, while TMP reaches 306.2 mmHg.

> ✦ DID YOU KNOW? **Changing Bags One at a Time**
>
> From the simulator: in the day-long challenge the worst error of the suggested configuration is 51.2 mL behind. The same configuration with the bags changed separately, not together, has a smaller worst error, 37.3 mL behind, and still passes, at 87.9 % of the dose against 89.6 % (Table 5.7).

5.7 Technology in Practice

The roller pumps, scales and pressure transducers of Figure 5.1 stand for the hardware of a machine for continuous kidney replacement. Figure 5.5 draws the loops of Section 5.5 with that hardware in place of the schematic's symbols.

What Measures It

The balance is measured by weight. Each bag hangs on a scale, and the weight gained by the effluent bag less the weight lost by the supply bags gives the net removal. Four transducers read the access,

Table 5.7 Challenge Configurations Over 24 h. Errors Are the Worst and the Final Deviation of the True Running Total

Configuration	Worst (mL)	Final (mL)	Dose (%)	TMP (mmHg)	Pass
as shipped	−893.0	−893.0	81.3	283.8	no
suggested[a]	−51.2	−24.1	89.6	205.5	yes
suggested, pump count	−979.3	−979.3	89.7	202.3	no
suggested, no supervision	−157.7	−134.8	89.5	202.6	no
suggested, no catch-up	−161.7	−157.5	89.7	205.9	no
suggested, no protection	−51.2	−27.3	92.2	306.2	no
suggested, no dose tracking	−49.9	−31.9	81.2	202.1	no
suggested, bags separately	−37.3	+27.4	87.9	204.3	yes

[a] scales, supervision, catch-up, protection, dose tracking, bags together

pre-filter, return and effluent pressures, and the transmembrane pressure is formed from three of them by the same combination that clinical studies of filter clotting use when they read the pressures a machine has recorded [18]. The chapter's scales follow Equation (5.9), with 2 g of correlated noise, a 250 g tare, a 1 g resolution and a bias that can drift, and its transducers read through a 1 s lag.

> **TECHNOLOGY Sensor: Bag Scales and Gravimetric Balancing**
>
> A machine for continuous therapy keeps its fluid balance by weight, by volume in balancing chambers, by flow meters, or by a combination. Weighing with one or more scales is the most common method, because it is the most reliable over long treatments. The effluent and the supply fluids are weighed continuously, the weight stands in for the flow, and the software adjusts the pump rates by servo feedback when prescribed and actual values disagree. Its weak points are the capacity of the scales, user errors and disturbances from the surroundings. Balancing by volume collects systematic, cumulative errors over a long treatment, since it lacks that continuous feedback [19]. Conventional machines control the balance to about ±30 mL/h, and one machine designed for newborns reports an ultrafiltration error of 1 g/h [20].

What Moves It

The final elements are four roller pumps, each turned by a drive under a speed loop. In the chapter a pump's delivery is its stroke times its speed, less the loading error of its tubing segment, its wear and its under-filling at a suction, as in Equation (5.1). The encoder on each pump counts revolutions exactly, and that count is the second, independent measurement of every flow that the scale supervision of Section 5.5 compares with the scales.

> **TECHNOLOGY Final Element: Roller Pumps on Tubing Segments**
>
> Every flow of the circuit is set by a pump. The blood pump sets the blood flow, the dialysate and replacement pumps feed their fluids, and the effluent pump sets the total fluid removed from the filter. Each works on a segment of tubing held between its rotor and stator, and pressure sensors watch the line before the blood pump, the line before the filter and the effluent compartment [19]. A blood pump of the usual kind has two rollers set 180° apart that squeeze a flexible tube. With a small catheter the circuit pressures must rise in proportion to keep the flow, and past a safe threshold the machine alarms stop the treatment [20]. United States regulation classes the delivery machine that pumps the blood and dialysate and controls the removal as a Class II device [21].

Where the Controller Runs

The controller is part of the delivery machine, which in United States regulation carries an ultrafiltration controller and mechanisms that monitor or control the fluid balance [21]. Its software reads

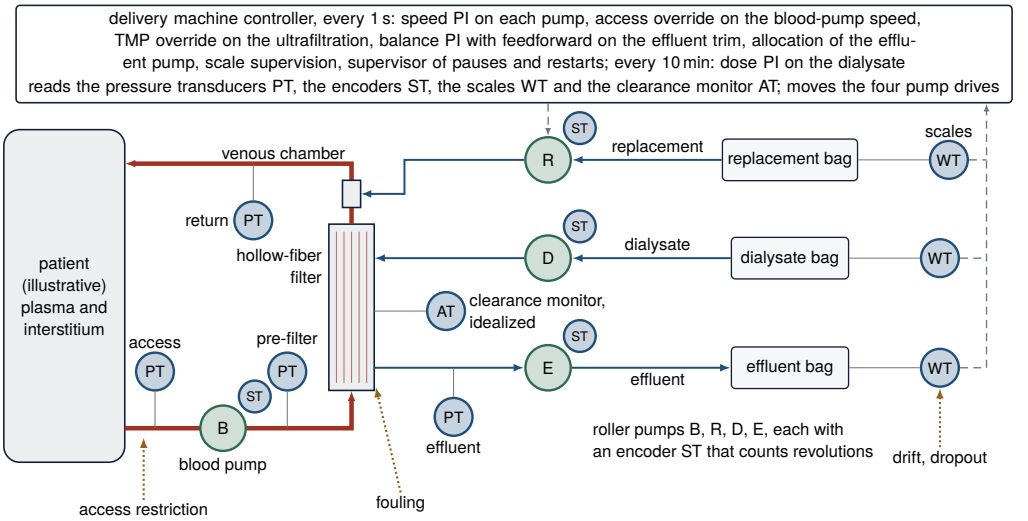


Figure 5.5 The loops of Section 5.5 with the hardware of a machine for continuous kidney replacement. Four roller pumps are the final elements: the blood pump B, and the replacement pump R, dialysate pump D and effluent pump E on the fluid side. The machine reads pressure transducers PT on the access, pre-filter, return and effluent lines, an encoder ST on every pump, a scale WT for every bag and, in the chapter only, an idealized clearance monitor AT. Thick red arrows carry blood, thin blue arrows the fluids, dashed lines the signals, dotted arrows the disturbances. The scales and the encoders measure the same flows twice, which is the redundancy the scale supervision uses.

the scales continuously and adjusts the pump rates by servo feedback [19], and the machine records its pressures and alarms for later review [2, 18]. No opened source gives the scan period of such a machine, so none is given here. The chapter's engine runs the speed loops, the two overrides and the balance loop once a second, the dose loop at each 10 min monitor sample, and the supervisor's bag changes, holds and restarts as a layer above them.

> **IN PRACTICE** **Watching Downtime and Alarms**
>
> Downtime, the share of treatment time with the therapy paused, is a quality indicator of continuous kidney replacement. In one cohort, access, return and filter pressure alarms were each associated with more downtime, and downtime widened the gap between prescribed and achieved fluid balance only in patients with six or more alarms a day [2]. A 2025 national guideline notes that the delivered dose often falls below the prescribed one, through interruptions, predilution and a less permeable membrane, so that 25 to 30 mL/kg/h is typically prescribed for a target of 20 to 25. It suggests a specialized or multidisciplinary team to lengthen filter life and reduce downtime [22].

What Is Modernizing It

Three developments bear on these loops. The first is data. Records from the machine and the health record can be screened by programs that alert clinicians to dangerous trends, and their suggestions can be fed back to the pumps, with or without a clinician's approval [9]. The second is a measurement of the patient. Bioimpedance and online hematocrit have been proposed for integration into the machine [9], and a multicenter randomized trial adjusted the ultrafiltration by bioelectrical impedance analysis [22]. Either would close a loop on the patient, which no loop of this chapter reads. The third is prediction. Early filter clotting is hard to detect, since the alarms sound only after the resistance has risen substantially, and resistances computed from the recorded pressures and flows have been proposed to predict it [18]. Machine learning has been tried for clotting, hypotension and alarm

management, though most such studies are retrospective and many lack external or prospective validation [23].

5.8 Limitations

The most basic limitation concerns purpose. The application is an educational equipment-control simulation of how a machine tracks a prescribed removal and dose with pumps, scales and pressure limits. Its patient model, its filter coefficients and every threshold it displays, the TMP limit and stop, the filtration-fraction ceiling, the access and pre-filter stop limits and the dose, are illustrative teaching values that have not had clinical review. It models no particular device, the prescriptions are scenario inputs that it never chooses, and nothing in it or in this chapter should be used to plan, adjust or judge the treatment of a patient.

OPEN CHALLENGE How Fast to Take the Water

The chapter takes the removal rate as a prescription it never chooses, and medicine has not settled that rate either. A 2025 guideline found no prospective intervention study comparing net ultrafiltration rates, rested its suggestion on retrospective studies and called for randomized trials [22]. A cohort of 973 patients in two centers then found the rate not independently associated with 28-day mortality once the achieved fluid balance was accounted for, while the balance itself was. Progress would be prospective studies that set patient-specific balance targets and use the rate as the means to reach them [24].

The technical limitations follow the assumptions. The patient has no blood pressure and no tolerance of removal, so the application cannot show a removal that the patient does not bear. Only urea is followed, and electrolytes and acid-base are absent. The membrane is lumped, its fouling is prescribed and the clearance monitor is idealized. The alarms are those the lessons need, and there is no air detector on the blood line, no blood-leak detector and no return-pressure alarm. The removal-unreachable report is implemented but no shipped scenario triggers it. Repeatability is itself unrealistic, since a fixed step and a fixed seed give the same day on every computer.

OPEN CHALLENGE Measuring the Dose Delivered

Dose estimates from the flows, the effluent or the pump settings, are easy at the bedside but often overestimate the dose found from urea clearance. Access dysfunction, recirculation, problems with anticoagulation and progressive fouling of the membrane lower the delivered dose without showing in the flows. Urea has limits of its own as a marker, and most solutes cannot be measured in routine practice [25]. The clearance monitor of Section 5.5 is idealized for that reason. Progress would be marker solutes measured in real time, at low cost, at the bedside [25].

5.9 Conclusion

A continuous kidney replacement circuit controls a small difference of large flows, and the simulator of this chapter carries it as a running total built from four imperfect pumps and three scales while it records the truth separately. On the pump counts the tubing tolerances, wear and suction leave the patient 466.3 mL behind after a day, and on the scales the balance stays within 22.3 mL. The balance loop, a PI loop on an integrator whose step response comes within 3 % of its linear prediction, passes a 60 g/h scale drift to the patient, 304.3 mL in 6 h, while its own error stays within 28.4 mL. A check against the pump counts flags that drift in 16.0 min but never flags a steady 35 g/h drift below its tolerance, which leaves the patient 379.3 mL behind in 12 h. At the effluent pump's limit the gain of the dose loop changes sign, from 0.427 to -0.493, and a cap that keeps effluent headroom holds the dose at 27.8 against 21.4 mL/(kg h) over hours 5 to 8. The challenge moves from 893.0 mL short and 81.3 % of the dose to within 51.2 mL and 89.6 %. The patient model is illustrative and the clearance monitor idealized. A fuller model would add the hemodynamic tolerance of removal, electrolytes,

clearance inferred from samples and a population of patients. The simulator is a browser plant for learning that a loop holding its own measurement need not be holding the patient.

KEY POINTS

- The water a patient loses is a small difference of large pumped flows, so a small pump error grows large in it.
- On the pump counts the patient ends a day 466.3 mL behind, and on the scales the balance stays within 22.3 mL.
- A balance loop on a drifting scale keeps its own error within 28.4 mL while the patient falls 304.3 mL behind in 6 h. A check against the pump counts flags that drift in 16.0 min.
- At the effluent pump's limit the gain of the dose loop changes sign, and a cap that keeps effluent headroom holds the dose at 27.8 against 21.4 mL/kg/h.

Where else this applies. A running total that only an independent measurement can check returns in Volume 5, Chapter 13, and a weighed flow held while the product is not in Volume 12, Chapter 12.

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5.10 Exercises

5.1 ⌕ Concept check · 2 min

In scenario 5 the effluent scale starts to read 60 g/h high at 1 h, and scale supervision is off. The balance loop holds the running total measured by the scales on its target. What do the trends show over the next five hours?

(a) The measured error stays within a few tens of milliliters while the true removal falls behind by about the drift rate

(b) The measured error grows at about the drift rate, and the badge shows the band crossed after about an hour

(c) Both errors stay small, because the integral action of the loop cancels a constant bias of the scale

(d) The measured and true errors each grow at about half the drift rate, because the loop splits the bias

5.2 ⌕ Concept check · 2 min

In case 1 the pumps count 930 mL/h of dialysate and 1000 mL/h of replacement into the circuit and 2030 mL/h of effluent out of it, for a net removal of 100 mL/h. Each pump in turn is given a stroke error of 1 %. Which statement is right?

(a) Each pump moves the net removal by the same amount, one percent of the removal
(b) The effluent pump's error is about a fifth of the removal, twice the effect of either supply pump
(c) The errors of the two supply pumps cancel the effluent pump's error when all three are present
(d) The errors move the balance only during bag changes, when the scales are off the bags

5.3 Concept check · 2 min

In scenario 8 the dose loop raises the dialysate until the effluent pump runs at full speed. Every further milliliter of dialysate then displaces a milliliter of replacement, and the dose falls while the loop pushes harder. Which change keeps the dose from collapsing, however long the run?

(a) Raise the dose-loop gain so that the loop reaches the prescribed dose before the pump fills
(b) Raise the replacement prescription so that convection makes up the clearance lost to diffusion
(c) Cap the dialysate below the effluent rating so that the pump keeps headroom for the other flows
(d) Balance on the pump counts so that the effluent command is no longer trimmed by the scales

5.4 Theory and calculation · 6 min

Case 1 prescribes a removal of 100 mL/h with 930 mL/h of dialysate, 1000 mL/h of replacement and 2030 mL/h of effluent by pump count.

(a) Compute the amplification $Q_k/Q_{net,Rx}$ of a relative error in each pump.*Checkpoint:* 20.3, 10.0 and 9.3

(b) The effluent pump moves 15.6 L in 8 h. Predict how much the removal error by pump count at 8 h moves per percent of effluent stroke error, and compare with Table 5.3. *Checkpoint:* 156 mL per percent

(c) At reset the true removal falls 11.2 mL/h short of the pump count. Predict the deficit after 24 h if the shortfall stayed constant, compare with the reported 466.3 mL, and name the two terms of Equation (5.1) that make the difference.*Checkpoint:* 268.8 mL

5.5 Theory and calculation · 6 min

The balance loop sees an integrator, $V_{net,m}(s) = \Delta(s)/s$ in the trim Δ, and a PI controller with gain K_c and integral time τ_I.

(a) For $K_c = 2/\text{h}$ and $\tau_I = 1\,\text{h}$, find the closed-loop poles, the damping ratio and the natural frequency. *Checkpoint:* $-1 \pm \text{j}$ per hour, 0.707, 1.41 rad/h

(b) An unmeasured removal rate steps by 150 mL/h. Find the peak error and its time after the step.*Checkpoint:* 48.4 mL at 0.785 h

(c) Repeat (b) for $K_c = 4/\text{h}$ and the same integral time. *Checkpoint:* 27.6 mL at 0.5 h

5.6 Theory and calculation · 6 min

The plausibility check of Equation (5.13) compares each scale with its pump's calibrated count over a window of 20 min, with $q_{abs} = 20\,\text{mL/h}$ and $q_{rel} = 1\,\%$, and flags a scale after two consecutive checks above the tolerance, one check a minute. All fluids weigh 1.00 kg/L.

(a) Compute the tolerance for the effluent scale at 2030 mL/h, the dialysate scale at 930 mL/h and the replacement scale at 1000 mL/h. *Checkpoint:* 40.3, 29.3 and 30.0 mL/h

(b) What is the smallest steady drift of the effluent scale that the check can flag, and how does Table 5.4 bear this out? *Checkpoint:* about 40 g/h

(c) A drift of 60 g/h starts at t_0. While the window still reaches back before t_0, the disagreement is $\delta q = \beta\,(t - t_0)/W_c$. Find the earliest time after the onset at which it exceeds the tolerance, and the earliest flag. *Checkpoint:* 13.4 min, flag from about 14.4 min

5.7 App challenge · 7 min

Open scenario 5, Scale drift, at the default seed and the default speed of 300 ×. Make no change before the first run.

(a) Run to 6 h with Scale supervision off as shipped. Read the Worst removal error on the scorecard, within 3 mL, and the range of the measured error on the first trend. *Checkpoint:* −304 mL; measured within about ±30 mL

(b) Press Reset, set Scale supervision to On before pressing Run, and run to 6 h. Read the Worst removal error and the effluent pump card. *Checkpoint:* −13 mL, effluent scale IMPLAUSIBLE

(c) Run on to 12 h and read the Worst removal error again. Explain why it grows although the drifting scale has been replaced. *Checkpoint:* −37 mL

5.8 App challenge · 8 min

Open scenario 7, Sensor dropout, at the default seed and 300 ×. The effluent scale is silent for 20 min from 1 h, and Scale supervision is off as shipped. Read each value on the Worst removal error tile after a run to 2.5 h, within 3 mL.

(a) Run as shipped. Read the tile and the lowest measured error on the first trend.*Checkpoint:* +80 mL; measured about −650 mL

(b) Press the reload button. On the Gravimetric balance card open the advanced settings, set Largest effluent trim to 1000 mL/h, and run. Reload again, set it to 100 mL/h, and run. *Checkpoint:* +209 mL and +24 mL

(c) Reload, set Scale supervision to On, and run. *Checkpoint:* −7 mL

5.9 App challenge · 8 min

Open the challenge, A day on the circuit, at the default seed. Run each day to 24 h at 900 ×, about 1.6 min a run, and read the scorecard.

(a) Run the day as shipped. Read the Worst removal error, the Delivered dose in percent, the Highest TMP and the Highest FF, and say which criteria fail.*Checkpoint:* −893 mL, 81 %, 284 mmHg, 21.1 %

(b) Reload, press Reveal suggested tuning, and run. Read the same four tiles.*Checkpoint:* −51 mL, 90 %, 206 mmHg, 22.5 %

(c) Reload, reveal the suggested tuning, set Change bags together to Off, and run. Does the day still pass?*Checkpoint:* yes, −37 mL, 88 %

5.10 Application area · 4 min

In case 4 the access override holds the measured access pressure at −150 mmHg. The blood pump then counts 111.6 mL/min while 103.5 mL/min actually passes, and the machine's filtration fraction reads 24.0 % against a true 25.4 %. Explain the discrepancy, say why it matters for the filtration-fraction ceiling, and name two ways a machine could narrow it.

5.11 Application area · 4 min

In case 6 ten bag changes pause the fluid pumps for 37.2 min in 8 h at a prescribed removal of 150 mL/h. Estimate the removal lost if nothing is paid back, compare with the reported deficit, and say what catch-up and changing the bags together each do about it.

5.12 Review the agent · 5 min

An AI assistant reviews scenario 5 and writes: *The balance loop is too slow to reject the scale drift. Raise the Balance loop gain from 2 to 6/h and cut its integral time from 1 to 0.3 h, and the true*

removal error will stay inside the band. Find the flaw, and say how you would check the claim in the app.