Rideout_IndicatorDilution

This model simulates the human cardiovascular system. It simulates the following five scenarios:
(1) Healthy Heart (with recirculation)
(2) Ventricular Septal Defect (VSD)
(3) Atrial Septal Defect (ASD)
(4) Patent Ductus Arteriosus (PDA)
(5) Open Loop (no recirculation)

The model is made of 10 compartments which combine perfect mixing and delay compartments. The anatomy represented by each compartment, as well as its volume and time constant are shown in the table above. This is a simple model which ignores capillary diffusion. An indicator bolus is injected into the pulmonary artery (compartment 1) and recirculates through the compartments in modes 1 through 4. However, mode 5, which is non-physiological, includes clearance through the right ventricle (compartment 0) and no recirculation.

For closed loop modes, by conservation of mass, the total amount of indicator in all compartments is fixed. Initially we expect to see the concentration peak in each compartment with the accumulated delays as shown in the table. The Transport Time around the loop is 40 sec. We therefore expect all compartments to show a peak-to-peak delay of ~40 sec (as long as the second peak can be distinguished). Eventually, the indicator concentrations will flatten into a unified concentration in all compartments.

In a healthy heart (mode 1) there are no internal loops. Blood flows from compartment to compartment in series. Figure 1 shows indicator concentrations in a healthy heart as computed by the JSim model. Note, for example, the indicator concentration at the small arteries (compartment 5). It peaks at ~15 sec, and again at ~52 sec. The peak-to-peak of ~37 sec is close to 40 sec as we expect. Also note that all concentrations flatten at ~0.00006 mM, the amount of injected indicator (0.25 umol) divided by total volume (4000 ml).

In a diseased heart (modes 2 through 4), there is an additional inner loop which flows in parallel to the outer loop and disrupts normal function. One or more of the conditions VSD, ASD or PDA can exist, although in this model we study the effect of each separately.

Two septal defects are studied in this model. Ventricular Septal Defect (mode 2), includes a shunt flow, Fshunt, from the left ventricle (compartment 3) to the right ventricle (compartment 0). Atrial Septal Defect (mode 3), includes a shunt flow, Fshunt, from the left atrium (compartment 2) to the right atrium (compartment 7). Figure 2 shows indicator concentrations in a heart with VSD as computed by the JSim model. Note the early peak in C0 due the shunt. In general, the effect of VSD or ASD is to generate lower peaks that occur later in time as the VSD or ASD aperture increases.

Rideout uses the pulmonary flow as a constant reference (F = 100 ml/s by default). This poses a design limitation with VSD or ASD (modes 2 and 3). In both cases Fs = F - Fshunt and therefore Fshunt cannot exceed F. The code includes a protection mechanism which limits Fshunt to 99% of F in order to prevent near-zero or negative systemic flow.

Patent Ductus Arteriosus (mode 4), includes a shunt flow, Fshunt, from the aorta (compartment 4) to the pulmonary artery (compartment 1). In this mode the reference flow is the systemic flow, and there is no risk of negative flow. The pulmonary flow is F = Fs + Fshunt.

Open loop (mode 5) is a non-physiological form of the cardiovascular system in which the effects of recirculation are subtracted off in some way. The injected indicator will eventually be cleared from the system through compartment 0.

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Open loop is useful for studying the cardiac output and system volume.
By conservation of mass, the same amount of indicator will go through
each compartment. The total area under the indicator injection pulse,
as well as the area under each concentration curve (time integral at
infinity) is a constant. Since the default amount of injected indicator
is 0.25 umol and average flow is 100 ml/s, area default is 0.0025 mM*s.
In open loop, we also compute Transport Functions, Hn(t), which
represent the fraction of solute arriving at the output of compartment
n per unit of time. Finally, we compute Average Transit Times, TAn(t),
from compartment 1 to any compartment n. This is the first moment of
the corresponding Transport Function Hn(t). Multiplying the final
value of the Average Transit Time by the flow F0 yields the aggregate
volume of blood of the compartments 1 through n.
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Rideout_IndicatorDilution_Modified

This model simulates the human cardiovascular loop. It can simulate a healthy or deseased heart. Both Ventricular Septal Defect (VSD) and Atrial Septal Defect (ASD) are modeled. To simulate a healthy heart, set both shunt flows Fvsd and Fasd to zero. To investigate a deseased heart set either or both to a positive flow.

Unlike Rideout's original model (Rideout_IndicatorDilution), this model keeps the systemic flow (Fs) as constant reference. This is more analogous to real life and is needed to keep us alive. When the systemic flow Fs is held constant, intracardiac flows must rise when there is a shunt, and observations of pulmonary hypertension and heart failure are easy to understand.

The model is made of 10 compartments which combine perfect mixing and delay compartments. The anatomy represented by each compartment, as well as its volume and time constant are shown in the table above. This is a simple model which ignores capillary diffusion. An indicator bolus is injected into the pulmonary artery (compartment 1) and recirculates through the compartments. By conservation of mass, the total amount of indicator in all compartments is fixed. In a healthy heart there are no internal loops. Blood flows from compartment to compartment in series. We expect to see the concentration peak in each compartment with the accumulated delays as shown in the table. The Transport Time around the loop is 40 sec. We expect all compartments to show a peak-to-peak delay of ~40 sec (as long as the second peak can be distinguished). Indicator concentrations eventually flatten into a unified concentration in all compartments.

In a diseased heart, there is an additional inner loop which flows in parallel to the outer loop and disrupts normal function. Ventricular Septal Defect (VSD) includes a shunt from the left ventricle (compartment 3) to the right ventricle (compartment 0). Atrial Septal Defect (ASD) includes a shunt from the left atrium (compartment 2) to the right atrium (compartment 7). Figure 1 shows indicator dilution C0 when the VSD shunt aperture is swept from 0 to 60 ml/s, which is 60% of the default systemic reference flow (100 ml/s). Note that in VSD, an early peak in the right ventricle is visible and increases in amplitude as the shunt aperture is increased. ASD creates a similar effect, where indicator dilution C7 shows an early peak.

Rideout_PressureFlowLH

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This model simulates the left heart using Pressure-Flow-Volume (or
Pressure-Flow for short) modeling. This technique takes advantage of the
analogy to RC or RLC circuits as shown here:

Pressure-Flow-Volume             Electrical Circuit Equivalent
-------------------------------------------------------------------
F  Flow (ml/s)                   I  Current (Amper)
P  Pressure (mmHg or g/cm/s^2)   V  Voltage (Volt)
Q  Volume (ml)                   Q  Charge (Coulomb or Amper*s)
R  Resistance (g/cm^4/s)         R  Resistance (Ohm or Volt/Amper)
C  Compliance (cm^4*s^2/g)       C  Capacitance (Farad or s/Ohm)
L  Inertance (g/cm^4)            L  Inductance (Henry or s*Ohm)

Pressure-Flow components obey the same fundamental equations as
their RLC circuit equivalents, i.e.:

P = R * F                        V = R * I    (Eq. 1 or Ohm's Law)
P = L * F:t                      V = L * I:t  (Eq. 2)
P = Q / C                        V = Q / C    (Eq. 3)
F = Q:t                          I = Q:t      (Eq. 4)

Blood flows from the left atrium to the left ventricle via the mitral
valve and continues to the aorta via the aortic valve. The load consists
of systemic and capillary arteries. Atrial pressure (PAT) is assumed
fixed at 6 mmHg as is the central venous pressure (PSV) at 3 mmHg. These
are equivalent to input and output electric potentials (voltages) which
drive the flow (current) through the left heart. Mitral and aortic
valves are modeled by limiting the flow to be positive and less than an
upper bound.

There are three separate modes implemented in this model, corresponding
to Rideout's ACSL programs:
(1) LH-PF-1: Rectangular Actuator
(2) LH-PF-2: Clipped Half-Sine
(3) LH-PF-3: Skewed Half-Sine and Inertance

LH-PF-1 and LH-PF-2 use a basic RC circuit; LH-PF-3 uses an RLC circuit
with additional components. The actuating signal for all modes is a
variable periodic left ventricular stiffness SLV (SLV = 1/CLV) which
simulates a contracting and expanding left ventricle during the cardiac
cycle. LH-PF-1 uses a rectangular SLV stiffness SLV waveform. LH-PF-2
uses a clipped half-sine. LH-PF-3 uses a clipped half-sine with a second
harmonic term which adds some skew to the waveform.

The figures show some results when running the JSim model in LH-PF-3.
Figure 1 is the ventricular stiffness, SLV. Stiffness changes between a
maximum, SLS during systole which lasts 0.3 sec, and a minimum, SLD
during diastole which last 0.5 sec. The pulse is periodic with a period
of 0.8 sec (corresponding to 75 beats/min heart pulse). Four heart beats
are simulated for a total run time of 3.2 sec. The left ventricle
volume, QLV expands during diastole and contracts during systole.

Figure 2 shows the locus (path) of the left ventricular pressure, PLV,
plotted versus volume, QLV. Values in this model were chosen so that the
end-diastolic volume (EDV) is 120 ml and the end-systolic volume (ESV)
is 48 ml. These volumes can be derived from Equation 3, P = Q / C, or
Q = P / S and are typical in a healthy human adult. The ejection
fraction (EDV - ESV) / EDV is 60% and stroke volume EDV - ESV is 72 ml.
At a heart pulse of 0.8 sec, this corresponds to a cardiac output of
90 ml/sec (5.4 liter/min). The locus is traversed in counterclockwise
direction. Note that it lies between the lines whose slopes are the
diastolic (minimum) muscle stiffness SLD and the systolic (maximum)
stiffness, SLS.

Figure 3 shows various flows in the model. Note that FAT, the atrial
flow rises then decays exponentially at diastole. This represents the
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diastolic inflow, which occurs when the walls of the left ventricle are relaxed and flow can pass through the mitral valve. As this chamber fills and expands, FAT decays. FLV, FAO and FSA have reverse activity, peaking and decaying during systole. FLV is the ventricular outflow and is limited by the aortic valve. FCA, capillary arteries flow, represents the systemic load and is much smaller and less pulsatile.

Figure 4 shows various pressures in the model. PLV, the ventricular pressure increases during systole, closing the mitral valve, and with further increase opens the aortic valve. Aortic pressure PAO follows ventricular pressure PLV during systole, while load pressure PCA is much more rounded. During diastole, PAO and PCA nearly coincide. The slow exponential decay has a time constant TCA = RCA * CCA of 2.53 sec. Note that some oscillations are visible in PAO starting at the end of systole.

Rideout_PressureFlow0

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This model simulates the complete CV loop using Pressure-Flow modeling.
The model is uncontrolled, i.e. does not include baroreceptor sensor
connections to the central nervous system (CNS). Also, it does not
consider blood diffusion from and to tissue. The model takes advantage
of the analogy to RLC circuits as shown below:

Pressure-Flow-Volume             Electrical Circuit Equivalent
-------------------------------------------------------------------
F  Flow (ml/s)                   I  Current (Amper)
P  Pressure (mmHg or g/cm/s^2)   V  Voltage (Volt)
Q  Volume (ml)                   Q  Charge (Coulomb or Amper*s)
R  Resistance (g/cm^4/s)         R  Resistance (Ohm or Volt/Amper)
C  Compliance (cm^4*s^2/g)       C  Capacitance (Farad or s/Ohm)
L  Inertance (g/cm^4)            L  Inductance (Henry or s*Ohm)

Pressure-Flow components obey the same fundamental equations as
their RLC circuit equivalents, i.e.:

P = R * F                        V = R * I    (Eq. 1 or Ohm's Law)
P = L * F:t                      V = L * I:t  (Eq. 2)
P = Q / C                        V = Q / C    (Eq. 3)
F = Q:t                          I = Q:t      (Eq. 4)

The right and left heart pathways are modeled with an RLC circuit
similar to that used in Rideout's left heart model LH-PF-2 mode, ported
to JSim in model Rideout_PressureFlowLH. Alternatively, this model may
be described as a compartmental representation of mass-transport, where
each of the 12 compliances in the RLC circuit, corresponds to a perfect
mixing chamber.

Circuit segment 0 corresponds to the right ventricle. Segments 1, 2 and
3 represent the pulmonary circulation. Segment 4 corresponds to the left
ventricle. Oxygenated blood leaves the left ventricle and flows through
ascending aorta (segment 5) and descending aorta (segment 6). There are
two major arterial systemic paths: segment 7, which represent the
abdominal aorta and segment 11, which represents the upper body arteries
(fed through the common carotid). The abdominal aorta bifurcates into
the common iliac arteries which feed internal organs and leg arteries
(segment 8). The systemic venous system returns deoxygenated blood from
the legs (segment 9), upper body (segment 12) and internal organs
(segment 13). These veins drain into the vena cava (segment 10) and
empty into the right atrium.

Unlike Rideout_PressureFlowLH, this model has no fixed potentials
(pressures). Both the left and right ventricles have time-varying
compliance or stiffness (S = 1 / C). The actuating signals are periodic
half-sines S0 and S4 for the right and left ventricles stiffness
respectively. Systole and diastole durations are both 0.4 sec, so that
heart pulse is 0.8 sec (corresponding to 75 beats/min heart pulse). Peak
systolic stiffness is chosen such that it is approximately four times
larger in the left heart than in the right heart. Valves are simulated
using flow limiters.

This model may be used to study the effect of Ventricular Septal Defect
(VSD). This is done with an optional resistive shunt flow represented by
R40 = 1 / GD, where GD is the VSD conductance. When GD is zero, R40 is
infinite and there is no VSD flow.

Initial conditions in this model were chosen based on the steady state
at the end of diastole. Note that these values are slightly different
than the ones used in Rideout's ACSL program.

Figure 1 shows the left and right ventricular volumes, Q4 and Q0. As
seen, the stroke volume pumped out of each ventricle is approximately
90 ml. At a heart pulse of 0.8 sec, this corresponds to a cardiac
output of 112 ml/sec (6.7 liter/min). End-systolic volume (ESV) and
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end-diastolic volume (EDV) are similar in each ventricle, around 160 ml.
This means that the ejection fraction (EDV - ESV) / EDV is 56%.

Figure 2 shows the right and left ventricular pressures, P0 and P4. As
seen, peak pressure is about four times larger in the left ventricle.

Figure 3 shows the loci on the Pressure-Volume plane of the right and
left ventricles. Each systole-diastole cycle corresponds to one full
counterclockwise traversal of the loci.

Figure 4 shows the effect of introducing VSD flow. In this case, the
area under the aortic flow and VSD shunt flow are approximately the
same. Thus, about half of the left ventricle output is returned to
the right ventricle through the defect.

Rideout_PressureFlow1

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This model simulates the complete CV loop using Pressure-Flow modeling.
The model is uncontrolled, i.e. does not include baroreceptor sensor
connections to the central nervous system (CNS). Also, it does not
consider blood diffusion from and to tissue. The model takes advantage
of the analogy to RLC circuits as shown below:

Pressure-Flow-Volume             Electrical Circuit Equivalent
-------------------------------------------------------------------
F  Flow (ml/s)                   I  Current (Amper)
P  Pressure (mmHg or g/cm/s^2)   V  Voltage (Volt)
Q  Volume (ml)                   Q  Charge (Coulomb or Amper*s)
R  Resistance (g/cm^4/s)         R  Resistance (Ohm or Volt/Amper)
C  Compliance (cm^4*s^2/g)       C  Capacitance (Farad or s/Ohm)
L  Inertance (g/cm^4)            L  Inductance (Henry or s*Ohm)

Pressure-Flow components obey the same fundamental equations as
their RLC circuit equivalents, i.e.:

P = R * F                        V = R * I    (Eq. 1 or Ohm's Law)
P = L * F:t                      V = L * I:t  (Eq. 2)
P = Q / C                        V = Q / C    (Eq. 3)
F = Q:t                          I = Q:t      (Eq. 4)

The venous segments of this model, both pulmonary and systemic, have a
structure much like the RLC circuit used in the artery in Rideout's left
heart model LH-PF-2, ported to JSim in model Rideout_PressureFlowLH. The
model has a single loop, but arteriovenous pathways may be easily added.

compliances are regarded as linear, but with unstressed volume (with
zero pressure). This is different than Rideout's CV loop model PF-0,
ported to JSim in model Rideout_PressureFlow0, which assumed no
unstressed volumes. In this model the inital condition in Eq. 4 above,
Q(0), consists of a stressed and unstressed components:

Q(0) = QU(0) + QS(0)

where QU = QU(0) is fixed. The pressure in Eq. 3 above, is obtained from
the stressed portion:

P = QS / C = (Q - QU) / C

Left and right ventricle stiffnesses SLV and SRV (S = 1 / C) in this
model use a periodic half-sine with a second harmonic added to shape the
actuator more realistically. Systole lasts 0.3 sec and diastole 0.5 sec,
such that the heart pulse is 0.8 sec (corresponding to 75 beats/min
heart pulse). Valves are simulated using flow limiters.

This model may be used to study the effect of myocardial infarction of
one or both ventricles introduced at a chosen time. It also allows study
of the effect of infusing or removing blood (bleeding) at the aorta at a
chosen time. These conditions are modeled with a JSim choice variable.

Figure 1 shows the ventricular volumes, QLV and QRV. Stroke volume
pumped out of each ventricle is 64 ml. At a heart pulse of 0.8 sec, this
corresponds to a output of 80 ml/sec (4.8 liter/min), which is
slightly low for an adult male.

Figure 2 shows the left ventricular locus on the Pressure-Volume plane.
Each systole-diastole cycle corresponds to one full counterclockwise
traversal of the loci. End-diastolic volume (EDV) is about 130 ml and
end-systolic volume (ESV) is around 66 ml. This means that the ejection
fraction (EDV - ESV) / EDV is about 50%.

Figure 3 shows the right ventricular and pulmonary artery pressures.
Figure 4 shows the left ventricular and aortic pressures. Figure 5 shows
the flow out of the first segments of the aorta (FA1) and left atrial
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flow (mitral valve or left ventricular inflow). Oscillatory decay can be observed after the aortic valve close.

The rest of the figures show the effect of a myocardial infarction. At time 3.2 sec, systolic stiffness in both ventricles has been reduced by 25%, as seen in figures 6 and 7. Figure 8 shows the ventricular volumes. Stroke volume is suddenly decreased at infarction time. A slow recovery begins to occur as atrial average volumes and pressures increase. The right ventricle end-diastolic volume, in particular, is seen increasing and as stated by Starling's Law, stroke volume rises. This demonstrates that the CV system is stable to a certain extent, even without CNS control.

Rideout_PressureFlowReg

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This Pressure-Flow model of the CV loop is regulated with baroreceptor
feedback. It is similar to Rideout's uncontrolled CV loop model PF-1,
ported to JSim in model Rideout_PressureFlow1. Despite the uncontrolled
nature of that model, the Frank-Starling mechanism ensures stability.
More robust stability, however, requires regulation or feedback.

The present model simulates one of the body's important homeostatic
mechanisms to maintain blood pressure. Known as baroreflex, it is a
negative feedback loop with connections to the central nervous system
(CNS). Pressure sensors called baroreceptors are located in the carotid
(aortic) arch. Afferent nerves carry signals from these sensors to the
CNS and efferent nerves carry signals back to the heart. There are two
branches of efferent nerves with opposing effects: sympathetic which
elevate blood pressure and parasympathetic which lower it.

Baroreceptor regulation can be described as follows:
* If blood pressure increases:
  - Baroreceptors activated
  - Parasympathetic nerves activated, sympathetic inactivated
  - Heart rate and strength of contraction decrease
  - Blood pressure drops (returns to normal)
* If blood pressure drops:
  - Baroreceptors inactivated
  - Parasympathetic nerves inactivated, sympathetic activated
  - Heart rate and strength of contraction increase
  - Blood pressure increases (returns to normal)

A JSim choice variable is used to switch between open-loop (without
baroreceptor feedback) and closed-loop (with feedback). Left and right
ventricle stiffnesses SLV and SRV (S = 1 / C) in this model use a
half-sine pulse with a second harmonic added to shape the actuator more
realistically. With open-loop, these waveforms are periodic, with
systole lasting TS = 0.3 sec and diastole 0.5 sec, such that the heart
period is TH = 0.8 sec (corresponding to 75 beats/min). Peak amplitude
is LS = 2500 g/cm^4/s^2 for the left ventricle and RS = 350 g/cm^4/s^2
for the right ventricle.

With closed-loop, a feedback signal Y is derived from PA3, corresponding
approximately to carotid pressure. The following steps are taken:
1) Average PA3 using a low-pass filter. Filtered output, PA3F, tracks
   PA3, but changes less rapidly.
2) Calculate an error signal, DPA3F, as the difference between PA3F and
   a constant reference PREF = 72 mmHg. Error signal may be positive or
   negative, according to whether the average carotid pressure is
   greater or less than PREF.
3) Scale the error signal with a constant gain KX = 0.2 1/mmHg. Scaled
   output, Z3, is dimensionless.
4) Apply an offset and limit the error Z3 to obtain the feedback signal
   Y. Offset is 1.0, lower limit is 0.1 and upper limit 1.9. Thus, Y is
   unity when there is no error and has a maximum allowed deviation of
   0.9 about unity.

The goal of the feedback loop is to maintain the feedback signal Y as
close as possible to unity. Y is used to control the systolic amplitudes
LS and RS, heart period TH and systolic period TS using these equations:

LS = LSI / (0.5 + 0.5 Y)   (Eq. 1)
RS = RSI / (0.5 + 0.5 Y)   (Eq. 2)

TH = 0.2  + 0.6 Y          (Eq. 3)
TS = 0.14 + 0.2 TH         (Eq. 4)

With the feedback at unity, these values default to the open-loop
values. An increase in Y, which might result from an increase in carotid
pressure, will tend to increase heart period (decrease heart rate) and
also decrease strengths of contraction of both ventricles. The result of
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these changes will tend to return arterial pressures, and thus cardiac output, back to their original values.

This model may be used to study the effect of myocardial infarction of one or both ventricles introduced at a chosen time. The infarction is simulated by adding negative values DLS and DRS to the systolic ventricle stiffnesses LS and RS respectively at a chosen time. The model also allows study of arterial infusion or venous bleeding.

The figures examine open and closed loop behavior with total run time of 20 sec. A sudden myocardial infarction is introduced at time TI = 11 sec in the left ventricle. By setting DLS = 625, stiffness drops 25% from approximately 2500 g/cm^4/s^2 to 1875 g/cm^4/s^2. Figures 1 and 2 are in open loop mode; the rest of the figures are closed loop.

Figure 1 shows ventricular volumes, QLV and QRV. Stroke volume from the left ventricle is decreased at infarction time from approx. 68 ml to 55 ml. A slow recovery begins to occur as average volumes and pressures increase. By Frank-Starling's Law, stroke volume rises and at t = 20 sec it is approx. 63 ml.

Figure 2 demonstrates that without true feedback, some variables cannot return to their pre-infarction value. The carotid pressure, for example, drops from an approx. average of 72.5 to 70.5 mmHg.

Figures 3 and 4 show the feedback signal Y and heart period TH in closed loop. Both plots show a dip at 11 sec. Since the error signal DPA3F is negative, Y drops below unity. From feedback equations 3 and 4 above, heart period and systole period drop with Y (heart rate increases). Within approx. 4 sec, Y returns to near unity and TH returns to the open loop stable value of 0.8 sec. Thus the feedback loop was able to compensates for the infarction relatively rapidly.

Figures 5 and 6 show the left ventricle systolic stiffness LS and right ventricle systolic stiffness RS. From feedback equations 1 and 2 above, both increase when Y drops below unity. The increase in contraction strength contributes to the recovery. As Y returns to unity, LS and RS stablize with means near 1875 g/cm^4/s^2 and 350 g/cm^4/s^2.

Figures 7 shows ventricular volumes QLV and QRV. It is similar to the open-loop result of Figure 1, but recovery is faster.

Figure 8 shows the filtered carotid pressure in closed-loop. Unlike the corresponding open-loop result of Figure 2, PA3F returns near its value before infarction, centered at the reference point of PREF = 72 mmHg.

Rideout_PressureFlowNP

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Nonpulsatile models are useful for pharmacokinetic studies which
typically have time constants in minutes. Ignoring nonlinearities, the
ventricular pressure-volume locus is counterclockwise between two
straight lines whose slopes are SD=1/CD and SS=1/CS. CD and CS are the
diastolic and maximum systolic compliances of the myocardium. If we
denote the end-diastolic pressure PED and end-systolic pressure PES,
then the stroke volume for the ventricle is:

QSV = PED * CD - PES * CS

Multiplying by the heart rate H, gives the average outflow:

F = (CD * H) * PED - (CS * H) * PES

Assuming a direct relationship between average atrial (Pat) pressure
and PED, and between average arterial (Part) pressure and PES, we get:

F = Gpre * Pat - Gafter * Part

Where Gpre and Gafter are the preload (prior to contraction) and
afterload (during ejection) conductances, and are inversely proportional
to the diastolic and systolic stiffness, respectively.

Denoting the left ventricle conductances G1 and G2, and the right
ventricle conductances G3 and G4, the following equations describe the
nonpulsatile ventricular flow:

FL = G1 * PL - G2 * PS
FR = G3 * PR - G4 * PP   (Eqs. 1)

The pressure drops over the systemic and pulmonary peripheral
resistances are given by Ohm's law:

PS - PR = RS * FS
PP - PL = RP * FP   (Eqs. 2)

Assuming fixed compliances and unstressed volume (under zero pressure)
for each of the four components:

PS = (QS - QSU) / CS
PR = (QR - QRU) / CR
PP = (QP - QPU) / CP
PL = (QL - QLU) / CL   (Eqs. 3)

Finally, integrating the flow through each component give the volume.

QS:t = FL - FS
QR:t = FS - FR + FI
QP:t = FR - FP
QL:t = FP - FL        (Eqs. 4)

Since this is a nonpulsatile model, the steady state is a unified
constant flow through the loop. It is interesting to model a sudden
change and the system response until a new steady state is obtained.
Two such conditions are modeled with a JSim choice variable:
(1) Infusion: Infusion flow pulse FI is added to systemic arteries
(2) Left Ventricular Stiffness Change: G2 is halved at time TCH

In infusion mode, a flow pulse FI is added to the systemic arteries
flow (second equation in Eqs. 4). The infusion pulse is rectangular,
starting at time TSTT = 1 sec and lasting WID = 2 seconds. In this
mode all conductances are fixed.

In Left Ventricular Stiffness Change mode, there is no infusion and
total blood volume is constant. Left ventricular afterload
conductance G2 is halved (left systolic stiffness is doubled) at
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TCH = 1 sec.

Figure 1 shows the flow pulse FI in infusion mode. Figure 2 shows an increase of 50 ml in total blood volume (integral of flow). Figure 3 shows the flows FR, FP, FL and FS in infusion mode. Figures 4 through 11 show the corresponding volumes and pressures.

Figure 12 shows the flows FR, FP, FL and FS in Left Ventricular Stiffness Change mode. Figures 13 through 16 show the corresponding pressures. The increase in systolic stiffness results in increased PS and QS (as evident from Eqs. 3). PR and QR also increase, but somewhat more slowly. As a result of these increases, the variables PP, QP, PL, and QL decrease (total blood volume must remain constant). All flows settle to the same increased value because of the stronger left heart.