Multiscale modeling of cardiac valve disease using cell-level signals to drive myocardial growth

Hossein Sharifi 1, Austin G. Wellette-Hunsucker 2, Charles K. Mann 1, Jonathan F. Wenk 1,3, Kenneth S. Campbell 2

1Department of Mechanical Engineering, University of Kentucky, Lexington, Kentucky, USA

2Department of Physiology & Division of Cardiovascular Medicine, University of Kentucky, Lexington, Kentucky, USA

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**\* Correspondence:**Kenneth S. Campbell  
k.s.campbell@uky.edu

Keywords: Multiscale modeling, Myocardial growth, Baroreflex, Concentric growth, Eccentric growth (Min.5-Max. 8)

Abstract

Multiscale models of the cardiovascular system are emerging as effective tools for investigating the mechanisms that drive ventricular growth and biological remodeling. Such models can be used to evaluate the effects of molecular-level mechanisms on organ-level function, which could provide new insights for improving patient care. PyMyoVent is a multi-scale computer model that simulates a left ventricle pumping blood around a systemic circulation by bridging from molecular to organ-level mechanisms. In previous work, we implemented a baroreflex control of arterial pressure by using feedback to regulate heart rate, intracellular Ca2+ dynamics, the molecular-level function of both the thick and the thin myofilaments, and vascular tone. In this paper, we extend PyMyoVent with concentric growth (wall thickening / thinning) and eccentric growth (chamber dilation / constriction) driven by cell and molecular-level signals. Specifically, concentric growth is controlled by the energy used by the cells for contraction (expressed as myosin ATPase normalized to myofibrillar volume) while eccentric growth responds to intracellular passive stress. The new framework reproduced clinical measures of LV growth in three types of valvular disease, namely aortic stenosis, aortic insufficiency, and mitral insufficiency. Furthermore, simulations for each valvular disorder regained LV size and function (reversal of growth) when the disease-mimicking perturbation was removed. In conclusion, the simulations suggest that myosin ATPase normalized to myofibrillar volume and intercellular passive stress can be used to drive concentric and eccentric growth in simulations of valve disease.

# Introduction

The heart is able to adapt its shape and size in response to pathological conditions, such as altered ventricular loading from valvular disease. This process is referred to as cardiac growth and remodeling (Frey and Olson, 2003; Pitoulis and Terracciano, 2020). Based on the ventricular geometry, there are two conventional types of growth namely concentric and eccentric growth. The former is defined by wall thickening and an increase in ventricular mass, due to the deposition of sarcomeres in parallel, with little or no change in the ventricular chamber size (Hill and Olson, 2008). The latter, however, is characterized by the addition of sarcomeres in series, which results in ventricular dilation and elevated ventricular mass with little or no change in the wall thickness (Hill and Olson, 2008). In general, cardiac growth initiates as an early adaptive response to valvular diseases, but it can progress to heart failure if the underlying cause is left unresolved (Hill and Olson, 2008; Shimizu and Minamino, 2016; Nakamura and Sadoshima, 2018).

Computer based models are providing new insights on the progression of cardiac growth and remodeling. Despite numerous studies that have developed mathematical formulations to represent these phenomena, the choice of driving stimulus for these growth laws is still up for debate. Conventionally, computational models of cardiac growth have utilized either myofiber stress (Rausch et al., 2011; Klepach et al., 2012), strain (Guterl et al., 2007; Kerckhoffs et al., 2012; Witzenburg and Holmes, 2018), or some combination of the two (Goktepe et al., 2010; Arts et al., 2012) as their driving signal. Rondanina and Bovendeerd (Rondanina and Bovendeerd, 2020a) tested four combinations of myofiber stress and strain driven laws, for both concentric and eccentric growth, and concluded that using at least one stress-driven law would predict more reliable growth. In another work, Mojumder et al. (Mojumder et al., 2021) showed that concentric growth of the LV, due to pressure overloading, correlates better with myofiber stress than stretch.

Though previous models have shown promising results, the underlying mechanisms that drive growth are more complex and are accompanied by perturbations at the molecular level (including signaling pathways, hormone levels, energy metabolism, etc.) that have not been thoroughly investigated. Therefore, the focus of computational modeling in cardiac growth is shifting from phenomenological models towards more realistic multi-scale mechanistic models that incorporate the effect of molecular/cellular events in order to drive growth (Sharifi et al., 2021a). For example, Yoshida et al. (Yoshida et al., 2020b) incorporated a network model of cellular level signaling pathways with a compartmental model of a rat heart to investigate cardiac growth in response to volume overloading and a surge in hormone levels during pregnancy. Their multi-scale model showed that most of the growth, especially during the first half of pregnancy, was due to a rise in progesterone (i.e. hormonal signal) and not from the volume overloading (i.e. mechanical signal). In another work, Estrada et al. (Estrada et al., 2021) coupled the effects of hormonal and mechanical signals in a finite element model of the left ventricle (LV) to predict hypertrophy in response to transverse aortic constriction (TAC). They concluded that the effect of hormonal inputs on the prediction of cardiac hypertrophy was larger than the mechanical stimulus.

Davis et al. (Davis et al., 2016) have suggested that cardiac growth correlates with the magnitude of tension developed by myofilament over time. Although this index has been informative in understanding of the underlying pathological mechanisms, it might not be the best driving signal. That is mainly because cells cannot differentiate between time scales to determine the tension-time integral, instead they all are continuously sensing all the changes in their environment. Tension-time integral developed by myofilaments, on the other hand, reflects the work done by the heart. Therefore, cellular level mechanisms that associate with elevated work demand of the heart such as energy can be the potential driving signal for concentric growth. In this manuscript, we hypothesize the role of energy used by cells to perform the adequate work in driving the concentric growth by using the myosin ATPase normalized to myofibrillar volume as the stimulus signal for concentric growth in response to valvular diseases.

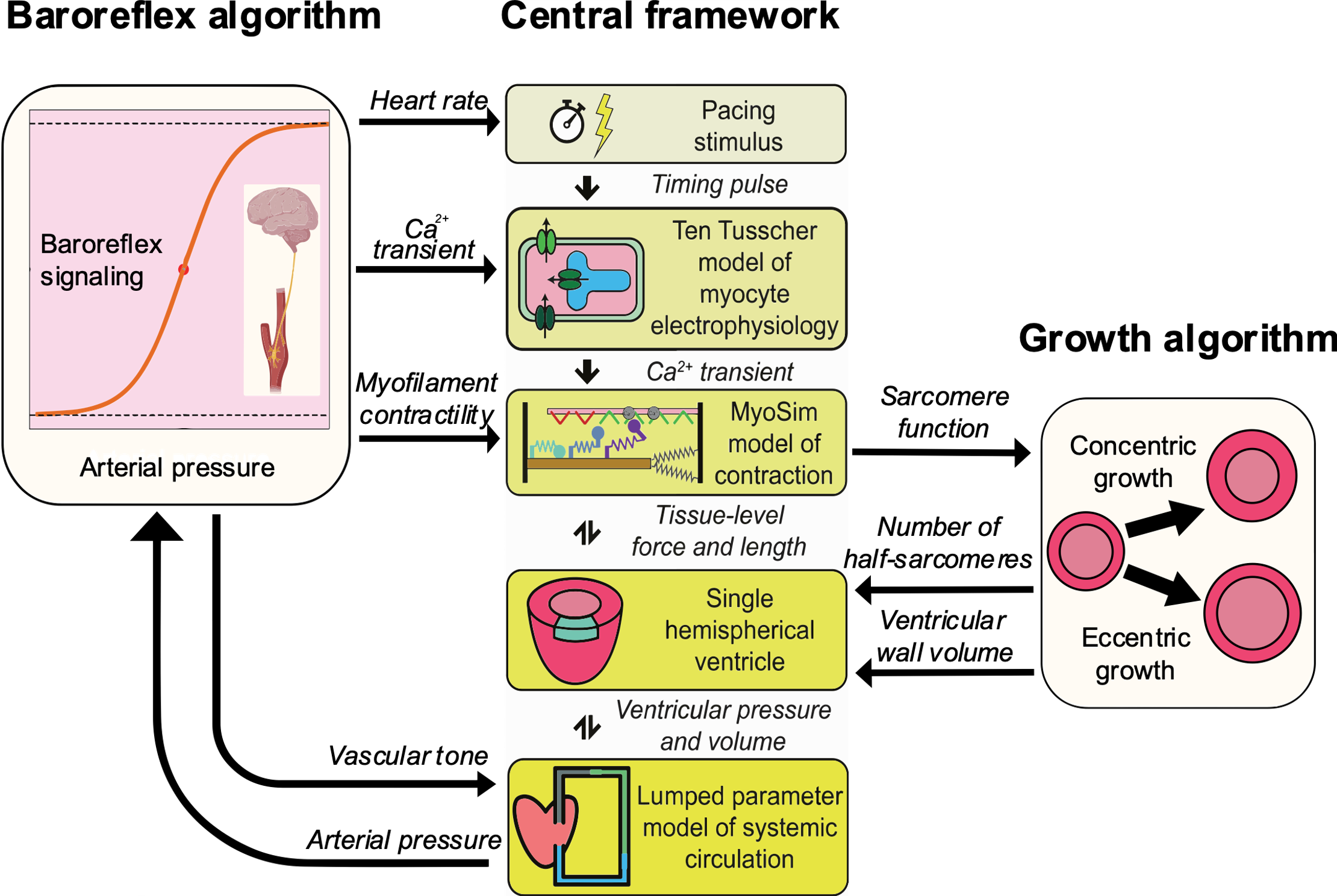
Eccentric growth, on the other hand, seems to be driven by a cell-level mechanism named mechanotransduction. Valvular diseases such as aortic insufficiency would lead to overstretching of sarcomeres that results into higher stress at the end of sarcomeres which in turn can be sensed by proteins located at this region (Knoll et al., 2002). Titin is a long protein that spans from Z disk to M line with an elastic structure within the I band that interacts with other proteins within the sarcomeric Z disk. The elastic portion of titin provides the passive stiffness of sarcomeres that stores strain energy during diastole (Lyon et al., 2015). Therefore, in current framework, we use the intracellular passive stress that reflects the stress within titin to drive the serial deposition of sarcomeres and eccentric growth of LV.

The objective of this manuscript is to investigate the role of cell-level signals in driving the concentric and eccentric LV growth in response to valvular diseases. To serve this purpose, we first implemented a growth module to a multiscale model of cardiovascular function named PyMyoVent (Campbell et al., 2020; Sharifi et al., 2021b) that uses myosin ATPase normalized to myofibrillar volume and intracellular passive stress in the half-sarcomeres to drive concentric and eccentric growth, respectively. Secondly, we tested the ability of the model to capture different types of LV growth in response to three kinds of valvular disorders, namely aortic stenosis, aortic insufficiency, and mitral insufficiency. Thirdly, we investigated the ability of current framework in regaining the LV size and function (reversal of growth) when the disease-mimicking perturbation was removed. Finally, we validated the model predictions for against clinical data that was compiled from the literature.

# Materials and Methods

## Overview

The current study extends our previous work (Sharifi et al., 2021b) by adding a module of LV growth to a multi-scale model of cardiovascular function named PyMyoVent. Figure 1 shows an overview of the PyMyoVent framework and illustrates how different modules communicate with each other. The original framework was published by Campbell et al. (Campbell et al., 2020), where they were able to show that a variation in model parameters (e.g. myosin rate constants) could change the system level parameters (e.g. end-systolic pressure-volume relationship). This original framework is essentially built on four main modules. First, a pacing stimulus is used to drive Ca2+ handling in the electrophysiology module. The contraction model called MyoSim (Campbell, 2014; Campbell et al., 2018) then uses calculated Ca2+ transient to predict the ventricular wall stress, which is then transformed into ventricular pressure via *Laplace’s Law*. Finally, a single hemispherical model of the LV pumps blood into a lumped parameter model of systemic circulation based on *Ohm’s law*. More details on these modules are provided in (Campbell et al., 2020).



**Figure** **1.** **Overview of the PyMyoVent framework.** The baroreflex algorithm regulates the arterial pressure towards a user-defined setpoint by modulating heart rate, intracellular Ca2+ transients, myofilament contractility, and vascular tone. The growth algorithm drives concentric growth (wall thickening / thinning) using myosin ATPase normalized by myofibrillar volume. Whereas the eccentric growth (chamber dilation / constriction) is driven using intracellular passive stress. Adapted from Campbell et al. (Campbell et al., 2020) and Sharifi et al. (Sharifi et al., 2021b).

## Baroreflex module

In our previous work (Sharifi et al., 2021b) we extended PyMyoVent (Campbell et al., 2020) by incorporating a module of the baroreflex feedback loop (Figure 1) to drive arterial pressure towards a user-defined setpoint. This work was essentially done by modulating the heart rate, intracellular Ca2+ transient, molecular-level function of both the thick and the thin myofilaments, and vascular tone. It was shown that the baroreflex algorithm was able to regulate arterial pressure towards setpoints ranging between ~30 mmHg to ~150 mmHg, as well as maintaining the arterial pressure under perturbed ventricular loading, such as acute blood loss or aortic stenosis. More details on the baroreflex module can be found in the previous work (Sharifi et al., 2021b).

## Growth module

The growth module algorithm was inspired by the underlying biology. *In vivo*, growth stimuli signals trigger a complex pathological downstream signaling pathway that promotes the cell growth and ventricular enlargement. In current model, a growth stimulus signal transduces into a normalized growth signal Ga,i that represents the net result of triggered upstream signals within the cell. The rate of change in Ga,i is defined as



where i represents the growth type (i.e. concentric or eccentric), ka,i is a rate constant and sets the speed at which Ga,i responds to a change in Si. Si,set is the homeostatic level (setpoint) for stimulus signal Si. During positive feedback, Ga,i tends towards one when Si is greater than Si,set and towards zero when Si is less than Si,set. *In vivo*, These bounds mimic the saturated levels of phosphorylation and dephosphorylation of underlying proteins by protein kinase.

The control signal Gc,i reflects the net results of downstream signals and governs how the effector parameters (i.e. wall volume or the number of half-sarcomeres) should respond to the normalized growth signal Ga,i. The rate of change in Gc,i is defined as



where γgrowth,i and γanti growth,i are rate constants that set the speed at which the effector parameters would grow or shrink according to Ga,i. For simplicity, rate constants (γgrowth,i and γanti growth,i)were chosen to have similar magnitudes but in the opposite directions.

Finally, the growth module links the control signals Gc,i to effector parameters as described in the following sections.

### Eccentric growth

In current model, eccentric growth was implemented by changing the number of serial half-sarcomeres (nhs) around the circumference of the left ventricle. The intracellular passive stress in the half-sarcomeres, *τ*passive, was considered as the stimulus signal (Secc) for eccentric growth. In current model, intracellular passive stress was modeled as



where xhs is the current length of the half-sarcomere, Lslack is the half-sarcomere length at which the passive stress is zero, L sets the curvature of the relationship, and is the scaling factor. The rate of change in the number of half-sarcomeres, nhs, is governed via equation , where Gc,ecc is the control signal for eccentric growth. According to equations - , number of half-sarcomeres (nhs) increases when Secc > Secc,set, but decreases when Secc < Secc,set.



### Concentric growth

Concentric growth was modeled by changing the left ventricular wall volume (Vwall) to mimic the parallel deposition of half-sarcomeres. Myosin ATPase normalized to myofibrillar volume (ATPase/Vmyofibrillar) was used as the stimulus signal for concentric growth (Scon). In current model, myosin ATPase normalized to myofibrillar volume is expressed as



where N0 is the number of myosin heads in a hypothetical half-sarcomere with a cross section of 1 m2, ∆G is the free energy produced by ATP hydrolysis (70 kJ mol-1), L0 is the reference length of half-sarcomere (1.1 μm), NA is Avogadro’s number (6.02 × 1023 mol-1), J4 is the detachment flux (s-1) of myosin heads from the force generating state (MFG) to the disordered relaxed state (MDRX), and x is the potential positions of myosin heads relative to the no-load position which varies from -10 to 10 nm with a resolution of 1 nm.

The rate of change in Vwall is defined via equation , which consists of two components. The first component, Gc,con, responds to the corresponding change in Scon, whereas the second component, Gc,ecc, incorporates the proportional change due to the eccentric growth (Pitoulis and Terracciano, 2020).



## Implementation and computer code

The code was written in Python using Numpy (Van der Walt et al., 2011) Scipy (Virtanen et al., 2020), and pandas (Reback et al., 2021) libraries. The source code and instructions on how to reproduce all figures shown in this manuscript are available at <https://campbell-muscle-lab.github.io/PyMyoVent/>.

Equations , , , and were discretized and implemented into the system of ordinary differential equations in PyMyoVent. Both nhs and Vwall were updated at each time-step. For simplicity, identical values were used for ka,ecc and ka,con. According to a simple sensitivity test (Figure S1), γgrowth,i and γanti growth,i rate constants only governed the speed of growth module at which reaches to the steady state, but not the magnitude of growth. Therefore, their values were chosen to manifest the LV growth in nearly less than a thousand heart-beat. The setpoints for both the concentric (Scon, set) and eccentric (Secc, set) growth laws were chosen to match the average value of the stimuli signals at steady state using default parameters.

## Simulations

### Baseline

As described in previous works with PyMyoVent (Campbell et al., 2020; Sharifi et al., 2021b) no data fitting was performed to optimize the model parameters. Instead, default parameters were selected to mimic the cardiovascular function of a healthy adult reported in the literature (Maceira et al., 2006; Petersen et al., 2017). All simulations shown in this manuscript started with the same assumptions using default model parameters. For example, total blood volume of the systemic circulation system was set to 4.5 liters and all simulations initiated with placing all stressed blood volume into veins. Similarly, in all simulations, baroreflex algorithm was activated at 20 s to maintain arterial pressure at setpoint of 90 mmHg when the simulation was at steady state using default parameters (Figure S2).

In this manuscript, most of figures show simulations that include at least hundreds of heart-beat. Therefore, pulsatile variables that vary remarkably during a cardiac cycle are shown with the envelope of extreme values over a cardiac cycle.



### Valvular disorders

Three types of valvular disorders, namely aortic stenosis, aortic insufficiency, and mitral insufficiency were simulated by applying the relevant perturbations to the baseline simulation.

According to *Poiseuille* equation, aortic valve area varies inversely with the aortic resistance. Hence, the stenotic aortic valve was modeled by increasing the aortic valve resistance, Rventricle, in equation that drives the blood flow from the left ventricle to aorta. In this manuscript, three levels of severity for aortic stenosis were modeled to represent the different stages of the disease according to American Heart Association (AHA) guidelines (Otto et al., 2021). For instance, a ~60 % reduction in the aortic valve area, from a mean value of 2.5 cm2 for healthy adults (Luszczak et al., 2012; Chin et al., 2014; Chin et al., 2017) to a mean value of 1 cm2 for patients with aortic stenosis (Spath et al., 2019; Everett et al., 2020) was equivalently mimicked by a 500 % increase in the aortic resistance. All simulated cases for aortic stenosis are summarized in Table 1.



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| --- | --- | --- |
| **Table** **1.** **Simulated levels of severity for aortic stenosis.** AS: Aortic Stenosis | | |
| % Increase in the aortic resistance | Equivalent % reduction in aortic valve area | Represented stage of disease according to AHA guidelines (Otto et al., 2021) |
| 250 % | 46.55  (From 2.50 cm2 to 1.33 cm2) | At risk of AS / Progressive AS |
| 500 % | 60  (From 2.50 cm2 to 1.00 cm2) | Asymptomatic severe AS |
| 750 % | 65  (From 2.50 cm2 to 0.86 cm2) | Symptomatic severe AS |

Aortic insufficiency was modeled by assigning a non-zero value to Gaorta in equation which essentially is the conductance (reciprocal of resistance) of the aortic valve. This would essentially allow a portion of blood volume in aorta to move backward to the left ventricle at diastole. American Heart Association (AHA) guidelines (Otto et al., 2021) categorize three levels of severity for aortic insufficiency based on the regurgitant volume, namely: mild (regurgitant volume < 30 ml beat-1), moderate (30 ml beat-1 < regurgitant volume < 59 ml beat-1), and severe AR (regurgitant volume > 60 ml beat-1). These levels of severity were simulated in this manuscript using the values for Gaorta shown in Table 2.

|  |  |  |
| --- | --- | --- |
| **Table** **2.** **Simulated different levels severity for aortic insufficiency.** | | |
| Gaorta | Equivalent regurgitant volume (ml beat-1) | Represented stage of disease according to AHA guidelines (Otto et al., 2021) |
| 5e-4 | 20 | Mild aortic insufficiency |
| 1e-3 | 40 | Moderate aortic insufficiency |
| 2e-3 | 70 | Severe aortic insufficiency |

Similarly, mitral insufficiency was simulated by giving a non-zero value to Gmitral in equation at which governs the blood flow between veins and the left ventricle. Three levels of severity were selected and simulated according to the regurgitant volume of 60 (ml beat-1) reported by American Heart Association guidelines (Otto et al., 2021). Simulated cases and represented stages of the disease are summarized in Table 3.



|  |  |  |
| --- | --- | --- |
| **Table** **3.** **Simulated different levels of severity for mitral insufficiency.** | | |
| Gmitral | Equivalent regurgitant volume (ml beat-1) | Represented stage of disease according to AHA guidelines (Otto et al., 2021) |
| 1e-3 | 30 | At risk of / Progressive mitral insufficiency |
| 2e-3 | 60 | Asymptomatic severe mitral insufficiency |
| 3e-3 | 80 | Symptomatic severe mitral insufficiency |







# Results

## Concentric growth in response to induced aortic stenosis

Figure 2 depicts a simulation of aortic stenosis in PyMyoVent framework. The initial transition in model responses is due to regulation of the arterial pressure towards the setpoint of 90 mmHg by the baroreflex module (Figure S1). At 50 s (first vertical dashed line from the left in all panels) the growth module was activated and then between the time points of 300 and 400 s (second and third vertical dashed lines) Raorta in equation was gradually increased by 500% from 20 to 120 (mm Hg L-1 s) to mimic a 60% reduction in the aortic valve area according to Table 2. The left-hand column demonstrates the response of the central framework shown in Figure 1 from molecular (e.g. fraction of binding sites on thin filament) to organ levels (e.g. LV cavity volume). The middle column reflects the modulation of reflex-sensitive parameters by the baroreflex module to control the arterial pressure. The right-hand column shows the properties relevant to the growth module such as the perturbed parameters at top panel and pertaining signals for concentric and eccentric growth in the following panels, respectively.

In cell-level, elevated aortic resistance initially increased the cell’s energy consumption for contraction (myosin ATPase normalized to myofibrillar volume, Scon) which led to an increase in Ga,con and Gc,con. The intracellular passive stress Secc, however, had only a subtle change and thus Ga,ecc and Gc,ecc did not alter much.

At organ-level, the change in the LV size was demonstrated by a ~30% increase in the ventricular wall volume (Vwall), ~8% reduction in the LV cavity volume at end-diastole, and almost no change in the LV cavity volume at end-systole. Ultimately, these changes together led to thickening of wall by ~21% and ~29% at end-systole and end-diastole, respectively, suggesting the occurrence of concentric growth. At the growth steady state, all cell-level signals (Ga,con, Gc,con, Ga,ecc, and Gc,ecc) normalized back to their homeostatic range as the LV geometry adapted the aortic stenosis condition.

Throughout the progression of growth, baroreflex module maintained the arterial pressure at the setpoint of 90 mmHg (middle column in Figure 2) by increasing the heart rate from ~63 to ~68 bpm, intracellular Ca2+ dynamics (kact and kSERCA), myofilament function (k1, k3, and kon), and vascular tone (Rarteriolar and Cveins).

Although the peak value of the stimulus signal for concentric growth Scon (myosin ATPase normalized to myofibrillar volume) appears higher at growth steady state than at baseline steady state, due to changes in heart rate and systolic duration caused by the baroreflex, the averaged value reaches the setpoint level for concentric growth (Figures S3-S4).

Diagram, schematic

Description automatically generated

**Figure** **2.** **Predicted concentric growth in response to aortic stenosis**. The left-hand column shows the responses of the central framework in PyMyoVent (Campbell et al., 2020) shown in Figure 1. The thin filament panel shows the fraction of actin binding sites in Noff and Non states. The thick filament panel shows the fraction of myosin heads in super-relaxed (MSRX), disordered relaxed (MDRX), and force-generating (MFG) states.

The middle column shows the baroreflex control of arterial pressure at setpoint of 90 mmHg. kact and kSERCA handle the intracellular Ca2+ dynamics, k1, k3, and kon handle the myofilament function, and Rarteriolar and Cveins handle the vascular tone.

The right-hand column shows the properties relevant to the growth module. Scon, Scon,set, Ga,con and Gc,con refer to the stimulus signal, setpoint, normalized growth signal, and control signal for concentric growth, respectively. Secc, Secc,set, Ga,ecc and Gc,ecc refer to the stimulus signal, setpoint, normalized growth signal, and control signal for eccentric growth, respectively.

The initial transition in all panels is due to baroreflex control of arterial pressure towards the setpoint of 90 mmHg (Figure S1). The growth module was activated at 50 s (first dashed vertical line from left on all panels) when the system was at steady state using default parameters. The system was gradually perturbed from 300 s to 400 s (second and third vertical dashed lines) by increasing Raorta (top panel in the right-hand column) by 500%.

## Concentric and eccentric growth in response to aortic insufficiency

Figure 3 shows an example of simulated aortic insufficiency condition. The simulation was performed similar to the one shown in Figure 2, but instead of changing Raorta, Gaorta in equation was increased from 0 to 1e-3 ([mmHg s]-1 L) to develop an insufficient aortic valve with a regurgitant volume of ~40 (ml beat-1) (Table 3).

In response to the induced insufficiency in the aortic valve, the initial rises in both stimuli signals (Scon and Secc) at cell-level drove the normalized growth signals (Ga,con and Ga,ecc) and hence elevated the control signals (Gc,con and Gc,ecc). Increased control signals were then recovered by increasing the ventricular wall volume and number of serial half-sarcomeres by ~45% and ~12%, respectively.

At organ-level, these changes resulted into the dilation of LV cavity (~38% at end-diastole and ~37% and at end-systole) and wall hypertrophy (~16% at both end-systole and end-diastole).

Although the baroreflex module maintained the arterial pressure setpoint at 90 mm Hg, arterial pressure became more pulsatile and changed from ~116/61 mm Hg to ~128/46 mm Hg.

Diagram, engineering drawing, schematic

Description automatically generated

**Figure** **3. Predicted concentric and eccentric growth in response to aortic insufficiency.** The panels are arranged similarly to those in Figure 2, except that aortic regurgitant volume is shown in place of aortic resistance in the right-hand column. The simulation shown in this figure was perturbed gradually (second and third vertical dashed lines) by increasing Gaorta in equation from 0 to 1e-3 ([mmHg s]-1 L) to induce an aortic regurgitant volume of ~40 ml (Table 3).

## Eccentric growth in response to mitral insufficiency

Figure 4 summarizes the model response to an example of mitral insufficiency simulation. The simulation started with the same setting described in Figures 2 and 3. Instead of changing Raorta or Gaorta, Gmitral in equation was increased from 0 to 2e-3 ([mmHg s]-1 L) to induce a regurgitant volume of ~60 (ml beat-1) through the mitral valve (Table 3).

At the cell-level, the insufficient mitral valve increased both stimuli signals for concentric (Scon) and eccentric (Secc) growth algorithms and consequently increased the relevant downstream control signals. The elevated control signals Gc,con and Gc,ecc were then re-normalized by driving the ventricular wall volume and number of serial half-sarcomeres to increase by ~50% and 17%, respectively.

At the organ level, these changes were manifested by dilation of LV cavity (~57% at end-diastole and ~68% at end-systole) and thickening of LV wall (~12% at end-diastole and 10% at end-systole). Such an excessive dilation in comparison to small thickening of wall describes the characteristics of eccentric growth in response to mitral insufficiency.

Due to baroreflex control of arterial pressure, along the subtle change in Ca2+ dynamics (kact and kSERCA), myofilament function (k1, k3, and kon), and vascular tone (Rarteriolar and Cveins), heart rate elevated from ~63 to ~66 bpm.

Although the peak value of intracellular passive stress at growth steady state seems different than at baseline steady state, due to changes in heart rate and diastolic duration, the averaged value reaches the setpoint level for eccentric growth (Figures S5-S6).

Diagram, schematic

Description automatically generated

**Figure** **4**. **Predicted eccentric growth in response to mitral insufficiency.** The panels are arranged similarly to those in Figure 2, except that mitral regurgitant volume is shown in place of aortic resistance in the right-hand column. The simulation shown in this figure was perturbed gradually (second and third vertical dashed lines) by increasing Gmitral in equation from 0 to 2e-3 ([mmHg s]-1 L) to induce a mitral regurgitant volume of ~60 ml (Table 3).

## Left ventricular pressure-volume loop relationship

Figure 5 illustrates the pressure-volume (PV) loops for simulated valvular disorders with different severities.

For aortic stenosis case (top panel in Figure 5), intensifying the severity of the disease resulted into higher peak systolic pressure. However, end-systolic LV volume remained unchanged and end-diastolic LV volume had a subtle decrease, which in turn led to a subtle reduction in stroke volume and ejection fraction. Also, the higher the aortic resistance in the model (Raorta in equation ), the larger the stroke work done by LV (the enclosed area by PV loop).

For aortic insufficiency case (middle panel in Figure 5), by increasing the level of insufficiency, the PV loop shifted to the right describing more dilation in the LV cavity volume. Higher regurgitant volume has also resulted into larger stroke volume and higher end-systolic pressure and thus larger stroke work done by LV. Furthermore, intensifying the severity of the disease resulted into more disturbance in the relaxation of LV.

Finally, increasing the level of mitral insufficiency led to more dilation of LV cavity by moving the PV loop to the right-hand side of the diagram (bottom panel in Figure 5). Increasing the severity of the diseases caused the stroke volume and thus the stroke work, along more disturbance in the LV relaxation, to increase. The peak systolic pressure, however, almost remained unchanged.

Diagram

Description automatically generated

**Figure** **5**. **Simulated** **left ventricular pressure-volume (PV) loop relationship for three types of valvular dysfunction with different levels of severity.** Baseline loop refers to the steady state response before applying any disease-mimicking perturbation. The other loops refer to final steady state solution after applying the relevant perturbation.

## LV recovery after removal of the overloading condition

In the next attempt, model was tested by removing the disease-mimicking perturbations when LV was in the growth steady state. Figure 6 depicts the reversal of LV growth when the underlying perturbations in Figures 2-4 were removed. Each column represents a simulated valvular disease. All three cases were started and perturbed exactly as shown in the original Figures (2-4). At 900 s (forth vertical line on all panels) the underlying perturbations were gradually lifted. For instance, aortic resistance was reduced from 120 to the default value of 20 (mm Hg L-1 s) for the aortic stenosis case. In all cases, LV dimensions (Fig 6) and function (Figures S7-S9) were fully regained to their homeostatic range once the underlying perturbation was lifted.

Diagram

Description automatically generated

**Figure** **6. Reversal of LV growth in response to removal of valvular diseases.** Each column summarizes the simulated response to removal of a valvular disease, where the left-hand column shows aortic stenosis, middle column shows aortic insufficiency, and right-hand column shows mitral insufficiency.On all panels, first vertical line reflects the activation of growth module. Second and third vertical lines demonstrate when the disease-mimicking perturbations were applied. Fourth and fifth vertical lines show when the underlying perturbations were removed.

## Importance of baroreflex control of arterial pressure

The effect of baroreflex control of arterial pressure was evaluated by redoing the simulations in Figures 2-4 with having the baroreflex algorithm deactivated. Simulations started with similar initial conditions shown in Figures 2-4, except the baroreflex algorithm was deactivated at 200s. Figure 7 demonstrates the effects of baroreflex control of arterial pressure on a selected group of model variables from the growth steady state.

For aortic stenosis case, arterial pressure dropped from ~113/64 mmHg, under control of baroreflex, to ~98/53 mmHg with no reflex control. LV end-systolic pressure also reduced from ~172 to ~146 mmHg. Consequently, due to altered hemodynamics, growth algorithm’s predictions for LV size reduced by ~11%, ~20%, and ~24% for LV end-diastolic volume, LV end-systolic volume, and LV wall volume, respectively.

For insufficient aortic valve simulation, the retrograde aortic blood flow did not change the arterial pressure in comparison to the case with baroreflex control, and thus the growth algorithm’s predictions for LV size remained nearly unchanged.

The simulation with insufficient mitral valve, resulted into a drop in arterial pressure from ~119/62 mmHg to ~109/54 mmHg as well as a reduction in LV end-systolic pressure from ~124 to ~114 mmHg. Additionally, LV cavity volume at end-diastole and end-systole along the wall volume were reduced by ~7%, ~14% and ~17%, respectively. More information regarding the full simulations is shown in Figures S5-S7.

Ultimately, in the absence of baroreflex algorithm, both the disease mimicking perturbations and modulation of growth module effectors led to alteration of the hemodynamics which in turn resulted into longer simulation time took by growth algorithm to reach to the steady state .

Chart

Description automatically generated

**Figure** **7. Effects of the baroreflex control of arterial pressure on simulated hemodynamics and growth module predictions.** Green bars reflect the results for growth steady state under control of baroreflex. Orange bars represent the variables at growth steady state without the control of baroreflex.

## Predicted results agreed with collected clinical data

To validate our model, the simulated results were compared with clinical data from the literature, which was acquired by cardiac magnetic resonance imaging (Table S1). Clinical data were categorized into four groups named control, patients with aortic stenosis, patients with aortic insufficiency, and patients with mitral insufficiency. For each category, measured data were collected from eight different studies shown in Table S1.

Ventricular dimensions were quantified with the LV end-diastolic volume index, LV end-systolic volume index, and LV mass index. Simulation results were normalized by an averaged body surface area of 1.9 m2 (Verbraecken et al., 2006; Lang et al., 2015) to match with the units of reported values in the literature. Statistical significances of the differences between the model predictions and clinical data were determined using two-sided equal variances t-tests.

Figure 8 shows model validation for predicting LV size with respect to the clinical data compiled from the literature (Table S1). Each column represents a diseased case, and each row represents a variable associated with LV size. For aortic stenosis case (left-hand column in Figure 8), by extending the severity of the disease, model predicted more increase in the LV mass index. However, LV volume index at both end-diastole and end-systole nearly remained unchanged. For the other two cases (insufficient aortic and mitral valves), predicted LV size parameters increased as the severity of insufficient valves increased. Calculated p-values from the statistical test, suggested that the model predictions for nearly all LV size parameters in all cases, except the LV mass index in response to mitral insufficiency, were not significantly different than the clinical data.

## Diagram, schematic Description automatically generated

**Figure** **8. Model validation for LV size in comparison to collected clinical data from the literature (Table S1).** Each column summarizes the model validation for a valvular disease (left, aortic stenosis; middle, aortic insufficiency; right, mitral insufficiency).In all panels, interquartile ranges for clinical data are shown with box plots in two groups of Control and Patient, whereas simulation results are shown with circle markers in two groups of Baseline (Sim) and Patient (Sim). LV end-diastolic volume index: LV end-diastolic volume normalized by the body surface area, LV end- systolic volume index: LV end-systolic volume normalized by the body surface area, LV mass index: LV myocardium mass normalized by the body surface area. ns (not significant): p ≤ 1.00e+00, \*: 1.00e-02 < p ≤ 5.00e-02

Systolic function was assessed with the LV stroke volume index and ejection fraction. Clinical data for systolic function was compiled similarly to LV size parameters. For studies (Table S1) that LV stroke volume index was not reported, the absolute difference between the reported LV volume index at end-diastole and end-systole was used instead.

Figure 9 summarizes the model validation for LV systolic function. According to Figure 9, by increasing the severity of aortic stenosis, model predicted small reduction in both the LV stroke volume index and ejection fraction. For the insufficient aortic valve, by increasing the severity of disease, predicted LV stroke volume index increased as well, but ejection fraction remained unchanged. For the mitral insufficiency condition, an increase in the regurgitant volume (RVmitral) resulted in higher predicted LV stroke volume index but lower ejection fraction. All predicted values for systolic function, except for the ejection fraction in response to mitral insufficiency, were not significantly different than the reported clinical data.

Diagram

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**Figure** **9. Model validation for LV systolic function in comparison to collected clinical data from the literature (Table S1.)** Figure panels are arranged like Figure 8. In all panels, interquartile ranges for clinical data are shown with box plots in two groups of Control and Patient, whereas simulation results are shown with circle markers in two groups of Baseline (Sim) and Patient (Sim). LV stroke volume index: LV stroke volume normalized by the body surface area. ns (not significant): p ≤ 1.00e+00, \*: 1.00e-02 < p ≤ 5.00e-02

# Discussion

This study extends an existing multiscale model of cardiovascular function by incorporating a growth module that simulates both concentric growth (wall thickening / thinning) and eccentric growth (chamber dilation / constriction). The simulation results showed that the new framework could predict the correct form of LV growth in response to three forms of valvular disease, namely aortic stenosis, aortic insufficiency, and mitral insufficiency. Model results were then validated with clinical data from the literature. Furthermore, simulations for each valvular disorder regained LV size and function (reversal of growth) when the disease-mimicking perturbation was removed.

## Role of myosin ATPase in driving concentric growth

In patients with chronic aortic stenosis, concentric growth is induced by the pressure overload experienced. Pressure overloading is a mechanical condition in which is characterized by an increase in the resistance of blood flow during LV systole. This condition reduces the shortening velocity of sarcomeres and thus increases the generated contractile force. Increased magnitude of developed force over time by myofilament has been suggested to have a significant relationship with concentric growth (Davis et al., 2016) which implies the total work performed by LV. Davis et al. (Davis et al., 2016) have also shown that none of the Ca2+ related parameters in their mice model correlated with the heart remodeling, suggesting probably as a weak driving signal for cardiac hypertrophy. Therefore, the metabolic demand of cells to meet such an elevated demand for performing adequate work in presence of pressure overloading seems to be the potential driver of concentric growth (cardiac hypertrophy).

Mitochondria are the main source of energy in eukaryotic cells and are abundant in high-energy-demanded organs like the heart. In healthy cardiomyocytes, mitochondria’s primary function is to meet the energy of the beating heart by producing ATP through oxidative phosphorylation. This makes up roughly 95% of the ATP production in the cardiomyocytes, with cross-bridge cycling of myosin heads consumes nearly 70% of ATP in the cell (Watkins et al., 2011). Considering the close relationship between workload and energy generation demand, concentric growth (cardiac hypertrophy) will inevitably lead to alterations in mitochondrial function, including mitochondrial dysfunction (Puddu et al., 2007; Green et al., 2011)

The increased in ATPase rate and increased ATP demand results in cardiomyocytes continuously synthesizing mitochondria to compensate for changes in energy demands and to remove damaged organelles, the process of which involves fusion and fission of existing mitochondria and separation of damaged ones for degradation (Iglewski et al., 2010). Too much mitophagy results in depletion of the mitochondrial population, while insufficient mitophagy will lead to damaged mitochondria accumulations (Iglewski et al., 2010), and an enviable shift in cardiac metabolism (Figure 10). It has been established that increased glucose utilization in hypertrophied hearts is a compensatory response to energy deficit caused by reduced fatty acid oxidation at a time of high energy demand for cardiac contraction (Tian et al., 2001; Luptak et al., 2005; Neubauer, 2007; Ritterhoff and Tian, 2017). Additionally, increased glycolysis has been strongly linked to cardiac hypertrophy (concentric growth), as well as an increased flux into ancillary pathways (Meerson et al., 1967). While preventing the switch of energy substrates in cardiomyocytes during pathological stimulation attenuates the influx of glucose into anabolic precursors and reduces hypertrophic growth (Ritterhoff et al., 2020). Therefore, the metabolic demand of cells to meet such an elevated demand for performing adequate work would increase in presence of pressure overloading and making the myosin ATPase an appropriate marker/driver of concentric growth (Figure 10)

With the metabolic switch from fatty acids to glucose is associated with an increase in anabolic metabolism, which provides glucose-derived aspartate for cellular hypertrophy. It has been demonstrated that a reliance on glucose for cardiomyocyte hypertrophy, a condition where energy requirement for contraction is removed (Ritterhoff and Tian, 2017; Ritterhoff et al., 2020). Lin28a is major regulator of pathological cardiac hypertrophy, which directly bound Pck2 mRNA to facilitate this metabolic repatterning in response to cardiac stress (Ma et al., 2019). Thus reveals a critical role of substrate switch for cell growth independent of energy demand. Lin28a enhances glucose uptake via an increase in insulin-PI3K-mTOR signaling (Zhang et al., 2014). Specifically, Lin28a increases IGF1 receptor, p-IRS-1, p-Akt, p-mTOR and p-p70s6k expression levels in cardiomyocytes (Zhu et al., 2011) (Figure 10). With the shift in metabolism, enhanced glucose up and increased IGF receptor expression (via Lin28a), the downstream signaling for stimulates pathological cardiac hypertrophy initiates.

It has been well established that increased IGF receptor expression activates PI3K (McMullen et al., 2004), which in turn can chronically activate Akt1 signaling. Chronic activation of the PI3K/AKT pathway occurs in cardiomyopathy. In vitro, the chronic activation of Akt1 gene expression can induce adaptive cardiac hypertrophy (Shiojima et al., 2005) by mTOR (Figure 10). The mammalian target of rapamycin (mTOR) pathway has been shown to be involved in the development of hypertrophic cardiomyopathy and is considered a therapeutic target for this disease (Lavandero et al., 1998). The Akt/mTOR pathway contributes significantly to the activation of mTORC1 during the development of cardiac hypertrophy (Volkers et al., 2013). Collectively linking increased metabolic state (myosin ATPase activity), fuel utilization shift (glucose utilization), and signaling for pathological hypertrophy (insulin-PI3K-Akt-mTOR signaling) (Figure 10).

In current framework, normalized myosin ATPase by myofibrillar volume has a relationship with detachment flux (J4) of myosin heads (equation ) which in turn is dependent on population of myosin heads in the force-generating state (MFG). During pressure overloading induced by valvular diseases, shortening velocity of half-sarcomeres decreases and thus due to less myosin heads being pulled off by strain the number of bound myosin heads in MFG increases. This essentially elevates the myosin ATPase reflecting higher energy demand for cells to produce enough contraction. In response to the increased stimulus signal for concentric growth (Scon), growth algorithm increases the normalized concentric growth signal Ga,con that reflects the net result of upstream signals. Elevated Ga,con, subsequently, drives the kinetics of control signal Gc,con reflecting the net result of downstream signals in cellular level in which modulates the parallel deposition of half-sarcomeres (concentric growth).

Diagram

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**Figure** **10. Role of energy demand (myosin ATPase normalized to myofibrillar volume) and intracellular passive stress (titin domains) in driving cardiac growth.**

Myosin ATPase and titin-domain architecture (N2A/N2BA-isoform of human cardiac muscle) laid out in a half-sarcomere for cardiac growth based signaling. Akt, protein serine/threonine kinase; ADP, adenosine diphosphate; ATP, adenosine triphosphate; CARP, cardiac-ankyrin-repeat-protein; DARP, diabetes-related ankyrin-repeat protein; ERK2, extracellular signal-regulated kinase-2; FHL2, four-and-a-half-LIM-domain protein; MAPK, mitogen-activated protein kinase; MAPK, mitogen-activated protein kinases; MARPs, muscle-ankyrin-repeat proteins; MEK1/2, MAP-ERK-kinase-1 and -2; MLP, muscle LIM protein; mTOR, mammalian target of rapamycin; MURF2, muscle-specific RING finger proteins(-2); Nbr1, neighbor of BRCA1 gene-1; NFAT, nuclear factor of activated T cells; p62, sequestosome 1/p62; P70s6K, p70 S6 kinase; PDK1, phosphoinositide-dependent kinase-1; PI3K, phosphatidyl inositol-3-OH-kinase.

## Role of intracellular sarcomeric passive stress in driving eccentric growth

In patients with valvular diseases such as chronic mitral/aortic insufficiency, eccentric growth is induced by volume overload. Such a condition initially results into excessive diastolic filling of LV and thus overstretching of sarcomeres before any remodeling happens. Emerging evidence links titin to fundamental signaling pathways, such as those regulating protein quality control, hypertrophic gene expression, and stress sensing. Titin can thus be viewed as a crucial integrating element at the crossroads of myocyte signaling. The mechanical and mechano-signaling functions of the titin springs are variably tuned in health and disease, particularly in the heart by altering passive stiffness through titin-isoform switching, protein phosphorylation, and hypertrophic signaling.

In heart muscle, titin is expressed in two main isoforms: the N2B-isoform, which contains a short, stiff spring segment, and (variable) N2BA-isoforms, which contain longer springs and thus are more compliant (Figure 10) (Freiburg et al., 2000). Titin is a long protein that spans from the Z disk to M line with an elastic structure within the I band. This elastic behavior of titin within the I-band plays an essential role in generating passive stiffness of the sarcomere which store strain-energy during diastolic filling and recoil during systole.

Cardiac titin has some unique properties that arise from the co-expression of N2BA and N2B isoforms in the half-sarcomere to the presence of the N2-A domain in the middle of the spring segment. The N2-Bus is an additional extensible element in the cardiac titin spring, next to the Ig regions and the PEVK segment (Linke et al., 1999), but it is also involved in protein-protein interactions (Figure 10). PEVK knockouts have been shown to trigger diastolic dysfunction through cardiac hypertrophy, presumably by increasing the binding of FHL1 to the N2-Bus, thereby activating the N2-Bus-associated stress sensor (Granzier et al., 2009). The N2-Bus binds two isoforms of the four-and-a-half-LIM-domain protein, FHL1 (Sheikh et al., 2008) and FHL2 (Lange et al., 2002). Both FHL1 and FHL2 are transcriptional co-activators and interact with effector mitogen-activated protein kinases (MAPKs). FHL1 bound to the N2-Bus associates with ERK2 and MEK1/2, as well as Raf1, which is activated via stretch and increase passive stress (Sheikh et al., 2008) and may suppress ERK2 and MEK1/2. Decreased or absent ERK1/2 signaling induces myocyte lengthening and eccentric growth (Kehat et al., 2011). Thus, FHL1 is a component of the stretch sensor at the I band that acts to sense stretch to restrict or lock the range at which physiological sarcomere length can extend following stretch to scaffold stress-induced interactions of MAPK components at titin in order to mediate ensuing hypertrophic signaling, which can lead to pathological cardiac hypertrophy.

A unique sequence of M-band titin is linked to regulatory pathways of muscle growth through binding to four-and-a-half LIM-domain protein-2 (FHL2). FHL2 has been shown to sense cardiac strain is sensed the M band signaling complex with Nbr1 and p62 (Radke et al., 2019). This protein has numerous other interaction partners, including metabolic enzymes (Lange et al., 2002), and appears to be a transcriptional co-activator. M band titin has other various links to pathways of muscle-growth regulation, particularly through the interaction with MURFs proteins that can shuttle to the nucleus to alter muscle gene expression. The titin kinase domain controls muscle gene expression and protein turnover via association with the neighbor-of-BRCA1 gene-1 (nbr1) protein, which in turn signals to MURF2 via binding to p62. MURF2 activates hypertrophic genes in the nucleus, such as serum response factor (Lange et al., 2005).

Links between titin and hypertrophic signaling mechanisms lastly, but additionally sensed at the Z disk titin domain (FIGURE #). Binding of the extreme NH2-terminal titin Ig domains, Z1/Z2, to telethonin (Zou et al., 2006) also recruits a telethonin-ligand, muscle LIM protein (MLP), to the Z disk (Knoll et al., 2002; Knoll et al., 2010). MLP has also been detected in the I band (Arber et al., 1997), at costameres, and abundantly in the cytosol, as well as in the nucleus. Shuttling of MLP to the nucleus (Boateng et al., 2009) can activate transcriptional regulators and may enhance protein expression. MLP also binds to calcineurin, a protein phosphatase dephosphorylating nuclear factor of activated T cells (NFAT), which can thus translocate to the nucleus and induce a hypertrophic gene program (Samarel, 2008). This hypertrophic pathway is thought to be activated by stress or strain imposed onto the Z disk, but the exact mechanism of action and the role of titin’s NH2 terminus in it remain obscure.

Lastly, Ig domains at titin’s N2-A-domain interact with the three homologous muscle-ankyrin-repeat proteins (MARPs), cardiac-ankyrin-repeat protein (CARP), diabetes-related ankyrin-repeat protein (DARP), and ankyrin-repeat-domain protein-2 (Ankrd2) (Mayans et al., 1998; Witt et al., 2005), which in turn bind to myopalladin (Bang et al., 2001), an important actin-regulating protein (Otey et al., 2005) (Figure 10). Since members of the MARP family also associate with transcription factors kojic (Kojic et al., 2004), a role for MARPs as nuclear regulators of transcription is likely. Thus, via MARP-binding, the N2-A-domain of titin could be involved in hypertrophic signaling mechanisms.

Overall, detected mechanical stimuli in the form of passive stress is sensed by sarcomeric titin domains that trigger a cascade of downstream signals that ultimately ends up with upregulating of protein synthesis, sarcomere addition and myocardium growth.

In our model, intracellular passive stress has a nonlinear relationship with the half-sarcomere length (equation ). Thereby, simulated valvular diseases such as aortic insufficiency that initially increases the diastolic filling of LV and overstretches the half-sarcomeres would increase the intracellular passive stress. In response to such an elevated mechanical stimuli, growth algorithm increases the normalized eccentric growth signal Ga,ecc, which in turn drives the kinetics of control signal Gc,ecc (Figure 10). Ultimately, by addition of half-sarcomeres in series, the half-sarcomere length and associated passive stress along Ga,ecc and Gc,ecc re-normalize back to their homeostatic range.

## Comparison with existing models of LV growth

Although many other computational models of LV growth have been developed and shed light on the underlying mechanics of LV growth, they still have limitations that need to be addressed (Sharifi et al., 2021a). Some of these limitations are related to the assumptions used for the duration of the cardiac cycle and the representation of systolic function. For instance, some models (Goktepe et al., 2010; Klepach et al., 2012; Lee et al., 2015a) have only simulated LV growth during diastolic loading and neglected systolic behavior of myocardium during ejection. Other models (Kerckhoffs et al., 2012; Lee et al., 2016; Arumugam et al., 2019) investigated the mechanics of LV growth, performed under a full cardiac cycle, where the contractile function was simulated using phenomenological Hill-type models. Another group of works (Witzenburg and Holmes, 2018; Estrada et al., 2021) have used a time-varying elastance model of the ventricle to simulate full cardiac cycle. Rondanina and Bovendeerd (Rondanina and Bovendeerd, 2020a; b) recently investigated different combinations of mechanical growth stimuli where they used a one-fiber model of cardiac function. This model essentially related the mechanics of the LV at the organ level expressed in terms of LV pressure and volume to mechanics at the tissue level expressed as sarcomere stress and length (Bovendeerd et al., 2006).

The current framework, however, simulates LV growth under full cardiac cycle in which the contractile behavior of the LV is driven by a mechanistic model of half-sarcomeres that simulates the sliding of myofilaments based on the Huxley crossbridge formation (Huxley, 1957) at the molecular level. By modeling the mechanics of half-sarcomeres, we are able to study the effects of pathological processes at the molecular level and how they affect disease development at the organ level. Additionally, this framework could potentially be used to study the effects of pharmaceutical interventions for treating cardiac diseases.

The absence of a baroreflex feedback loop is another limitation of existing models (Sharifi et al., 2021a). In general, existing models are performed under constant heart rate with no mechanism to control the arterial pressure. Kerckhoffs et al. (Kerckhoffs et al., 2012) observed that the absence of hemodynamic feedback was the potential cause of mismatch between calculated peak LV pressure in their model and experimentally measured values. Rondanina and Bovendeerd (Rondanina and Bovendeerd, 2020b) showed that by implementing a model of hemodynamic feedback into their growth model, they could address the observed reduction in mean arterial pressure and cardiac output found in their prior work investigating valvular disorders (Rondanina and Bovendeerd, 2020a). Current framework uses a baroreflex feedback loop to maintain arterial pressure by modulating heart rate, intracellular Ca2+ transient, function of both myofilaments, and vascular tone. As shown in the current results (Figure 7), applying disease-mimicking perturbations (e.g. aortic stenosis or mitral insufficiency) can change the arterial pressure and thus the whole hemodynamics when the baroreflex control of arterial pressure was already deactivated. Altered hemodynamics can in turn result into different outcomes for growth algorithm that might not be accurate. Furthermore, coupled baroreflex feedback loop in current framework has reduced the required time in which simulations reach to steady state solution for growth algorithm. For instance, in simulation of aortic stenosis condition without the baroreflex control of arterial pressure, growth algorithm did not completely reach to steady state even by nearly doubling the amount of simulation time (Figure S6).

The reversal of cardiac growth is a favorable outcome of clinical interventions for dysfunctional valves, i.e., the ventricle returns to a normal size and shape. Although existing computational models have shown success in predicting the development of growth, many of them are challenged when trying to predict the reversal of growth (Sharifi et al., 2021a; Yoshida and Holmes, 2021). For example, Yoshida et al. (Yoshida et al., 2020a) investigated the regression of growth due to the removal of pressure overloading, while using the growth law developed by (Kerckhoffs et al., 2012). Although this growth law performed the best in capturing the development of LV growth, in comparison to seven other growth laws (Witzenburg and Holmes, 2017), it could not predict the reversal of growth. Yoshida et al. (Yoshida et al., 2020a) further suggested that using an evolving setpoint could potentially address the inability of existing models to predict the reversal of growth. Of the few works that have studied the reversal of growth, Lee et al. (Lee et al., 2015a) modified a previously developed eccentric growth law (Goktepe et al., 2010) and were able to capture the reversal of growth for a realistic LV geometry under certain types of loading. Arumugam et al. (Arumugam et al., 2019) extended their previous work (Lee et al., 2015a) and investigated the development of anisotropic growth in a biventricular model of the heart in response to mechanical dyssynchrony. Using maximum elastic myofiber stretch over a cardiac cycle as the sole stimulus signal of their growth law, their model demonstrated growth in the left ventricular chamber size and septal wall, but reversal of growth for the right ventricular chamber size and LV free wall.

Our model, however, completely regained the LV size and function once the underlying perturbations reflecting the valvular disorders were lifted. This ability of current framework is possibly due to two reasons. Firstly, PyMyoVent framework uses a mechanistic model of half-sarcomere to simulate the contractile behavior of myocardium that captures length-dependent activation, cooperativity between thick and thin filaments, and the strain-dependent behavior of cross-bridges (Campbell, 2014; Campbell et al., 2018). Such a model can accounts for the effects of any alteration in the ventricular loading on the force generation of half-sarcomere that other models are unable to do so. For instance, Yoshida et al. (Yoshida et al., 2020a) had to manually adjust the muscle contractility in their model to mimic the lower force production of myocardium due to removal of pressure overloading. In contrast, in our model, removal of the aortic stenosis condition would lower the hemodynamics impedance during LV systole, which in turn increases the shortening velocity of half-sarcomere due to higher strain in myosin heads. This event reduces the number of bound myosin heads in force-generating sate (MFG) and thus lowers the associated generated force in half-sarcomere that matches with the altered hemodynamics loading. Secondly, current framework is benefited from being coupled with the baroreflex feedback loop. Therefore, there is no need to manually adjust the circulatory parameters when the overloading is removed to match with realistic hemodynamics as Yoshida et al. (Yoshida et al., 2020a) did in their work. Instead, such a feedback loop controls the arterial pressure by modulating heart rate, intracellular Ca2+ transient, function of both myofilaments, and vascular tone. This is for the first time, to the best of our knowledge, that LV growth is being simulated with this level of sarcomere mechanics in molecular level while the arterial pressure is being controlled by a baroreflex feedback loop.

## Limitations and future perspectives

The limitations discussed in the previous works with PyMyoVent (Campbell et al., 2020; Sharifi et al., 2021b) are still applicable to the current framework. However, the following limitations are particularly related to the growth module added in this work. Firstly, the current model can only capture uniform changes in the ventricular size and dimensions. This is due to the simplified 1-D hemispherical geometry of the LV, which does not account for the complex torsional motion of the heart (Russel et al., 2009), longitudinal and transmural variation of contractile properties (Sharma et al., 2003), or the variation in myofibers orientations (Rodriguez-Cantano et al., 2019).

Secondly, the current framework can only quantify the cardiac growth (i.e. change in the ventricular size and dimension), but not the myofiber remodeling. Alterations in mechanical loading (Pitoulis and Terracciano, 2020; Washio et al., 2020), as well as mutant sarcomeres in familial cardiomyopathy (Watkins et al., 2011), can be accompanied by myofiber disarray and remodeling. However, the PyMyoVent framework assumes the half-sarcomeres, and thus the myofibers, are uniformly placed around the circumference of LV at base and their orientation remains unchanged during LV growth. Thirdly, the current study does not include the effect of fibrosis that is commonly observed in patient with aortic stenosis (Treibel et al., 2018).

# Conclusions

This work extends a multiscale model of cardiovascular function by incorporating a growth module that simulates both concentric (wall thickening / thinning) and eccentric (chamber dilation / constriction) growth. The new framework reproduced clinical measures of LV growth in three types of valvular disease, namely aortic stenosis, aortic insufficiency, and mitral insufficiency. Additionally, the new framework could fully regain the LV size and function (reversal of growth) when the disease-mimicking perturbation was removed. In conclusion, the results of this study suggest that myosin ATPase normalized to myofibrillar volume and intercellular passive stress can be used to drive concentric and eccentric growth in simulations of valve disease.

**Acknowledgements**

# Supported by NIH HL133359 to KSC and JFW, NIH 148785 and TR0001998 to KSC, and AHA TP135689 to KSC.

**Author contributions**

SH drafted the manuscript, wrote prototype versions of the code, ran the final simulations, created the figures, helped develop the website and GitHub repository, and ran prototype simulations. CKM helped with planning the structure of the manuscript and edited the manuscript. AGWH helped with his knowledge in cell signaling by drafting sections 4.1 and 4.2. JFW helped develop the model framework and edited the manuscript. KSC planned the overall project, developed the growth algorithm, wrote the final version of the code, and edited the manuscript.

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# Supplementary material

Multiscale modeling of cardiac valve disease using cell-level signals to drive myocardial growth

Hossein Sharifi 1, Austin G. Wellette-Hunsucker 2, Charles K. Mann 1, Jonathan F. Wenk 1,3, Kenneth S. Campbell 2

1Department of Mechanical Engineering, University of Kentucky, Lexington, Kentucky, USA

2Department of Physiology & Division of Cardiovascular Medicine, University of Kentucky, Lexington, Kentucky, USA

3Department of Surgery, University of Kentucky, Lexington, Kentucky, USA

**\* Correspondence:**Kenneth S. Campbell  
k.s.campbell@uky.edu

Keywords: Multiscale modeling, Myocardial growth, Baroreflex, Concentric growth, Eccentric growth (Min.5-Max. 8)

## Supplementary tables

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| --- | --- | --- | --- | --- | --- | --- | --- | --- | --- | --- | --- |
| **Table S****1.** List of studies with quantified clinical data for LV dimensions and systolic function acquired by cardiac magnetic resonance imaging. | | | | | | | | | | | |
| **Control volunteers** | | | **Patients with Aortic stenosis** | | | **Patients with Mitral insufficiency** | | | **Patients with Aortic insufficiency** | | |
| Study | Year | n | Study | Year | n | Study | Year | n | Study | Year | n |
| Lee et al. (Lee et al., 2020) | 2020 | 30 | Lee et al. (Lee et al., 2020) | 2020 | 191 | Liu et al. (Liu et al., 2020) | 2020 | 104 | Malahfji et al. (Malahfji et al., 2020) | 2021 | 392 |
| Spath et al. (Spath et al., 2019) | 2019 | 41 | Everett et al. (Everett et al., 2020) | 2020 | 440 | Seldrum et al. (Seldrum et al., 2019) | 2019 | 59 | Seldrum et al. (Seldrum et al., 2019) | 2019 | 29 |
| Seldrum et al. (Seldrum et al., 2019) | 2019 | 30 | Spath et al. (Spath et al., 2019) | 2019 | 159 | Bakkesstrom et al. (Bakkestrom et al., 2018) | 2018 | 46 | Geiger et al. (Geiger et al., 2018) | 2017 | 16 |
| Lee et al. (Lee et al., 2015b) | 2015 | 15 | Singh et al. (Singh et al., 2019) | 2019 | 174 | Polte et al. (Polte et al., 2017) | 2017 | 40 | Polte et al. (Polte et al., 2017) | 2017 | 38 |
| Edwards et al. (Edwards et al., 2014) | 2014 | 35 | Everett et al. (Everett et al., 2018) | 2018 | 61 | Myerson et al. (Myerson et al., 2016) | 2016 | 152 | Fairbairn et al. (Fairbairn et al., 2013) | 2013 | 50 |
| Chin et al. (Chin et al., 2014) | 2014 | 33 | Chin et al. (Chin et al., 2014) | 2014 | 133 | Edwards et al. (Edwards et al., 2014) | 2014 | 35 | Myerson et al. (Myerson et al., 2012) | 2012 | 158 |
| Barone-Rochette et al. (Barone-Rochette et al., 2013) | 2013 | 20 | Barone-Rochette et al. (Barone-Rochette et al., 2013) | 2013 | 128 | Schiros et al. (Schiros et al., 2012) | 2012 | 94 | Uretsky et al. (Uretsky et al., 2010) | 2010 | 34 |
| Schiros et al. (Schiros et al., 2012) | 2012 | 51 | Steadman et al. (Steadman et al., 2012) | 2012 | 41 | Uretsky et al. (Uretsky et al., 2010) | 2010 | 23 | Grotenhuis et al.(Grotenhuis et al., 2007) | 2007 | 20 |
| Data were reported as mean ± standard deviation (SD) or median (interquartile range). | | | | | | | | | | | |

## Supplementary figures

**Diagram

Description automatically generated**

**Figure S****1. Sensitivity of growth module to rate constants of control signals γgrowth,con and γgrowth,ecc.** For simplicity, γanti growth,i were chosen to have similar magnitudes as γgrowth,i but in the opposite directions, where i reflects the growth type. γ0,con and γ0,ecc are the rate constants values that were used in all simulations of this study. The growth module was activated at 50 s (first dashed vertical line from left on all panels) when the system was at steady state using default parameters. The system was gradually perturbed from 300 s to 400 s (second and third vertical dashed lines) by applying the underlying valvular disorders.

Chart, bar chart

Description automatically generated

**Figure S****2.** **Baseline steady state using default parameters under control of baroreflex feedback loop.** Left hand column shows the response of central framework in PyMyoVent (Campbell et al., 2020). Right hand column shows the response of baroreflex algorithm. Simulation started with using default parameters. At 20 s, baroreflex algorithm was activated to move arterial pressure towards the setpoint of 90 mmHg by modulating heart rate, intracellular Ca2+ dynamics (kact and kSERCA), myofilament function (k1, k3, and kon), and vascular tone (Rarteriolar and Cveins). Non and Noff refer to status of actin binding sites. MSRX, MDRX, and MFG represent the status of myosin heads on thick filament in the super-realxed, disordered-relaxed, and force generating states, respectively. Parteries and Pset refer to the actual arterial pressure and setpoint for the arterial pressure, respectively.

Diagram

Description automatically generated

**Figure S****3. Alteration in the averaged myosin ATPase normalized by myofibrillar volume (Mean Scon) with respect to the concentric growth setpoint (Scon,set).** Results are replicated from Figure 2. Scon is stimulus signal for concentric growth law. Ga,con and Gc,con are, normalized growth and control signals for concentric growth, respectively

Chart

Description automatically generated

**Figure S****4. Comparison in myosin ATPase normalized by myofibrillar volume profile over a cardiac cycle at baseline and growth states.**

**Diagram

Description automatically generated**

**Figure S****5. Alteration in the averaged intracellular passive stress (Mean Secc) with respect to the eccentric growth setpoint (Secc,set).** Results are replicated from Figure 4. Secc is stimulus signal for concentric growth law. Ga,ecc and Gc,ecc are, normalized growth and control signals for eccentric growth, respectively

Chart, line chart

Description automatically generated

**Figure S****6. Comparison in intracellular passive stress profile over a cardiac cycle at baseline and growth states.**

Diagram, engineering drawing, schematic

Description automatically generated

**Figure S****7. Predicted recovery of LV size and function in response to removed pressure overloading.** Similar arrangement for panels as in Figure 2. Growth module activated at 50 s when the simulation was at initial steady state. On all panels, first vertical line shows when the growth module is activated. Second and third vertical line demonstrate the onset and ending of the applied pressure overloading. Fourth and fifth vertical lines shows the onset and ending of the removed pressure overloading.

Diagram, engineering drawing

Description automatically generated

**Figure S****8. Predicted recovery of LV size and function in response to removed aortic insufficiency.** Similar arrangement for panels as in Figure 2. Growth module activated at 50 s when the simulation was at initial steady state. On all panels, first vertical line shows when the growth module is activated. Second and third vertical line demonstrate the onset and ending of the applied aortic insufficiency. Fourth and fifth vertical lines shows the onset and ending of the removed aortic insufficiency.

Diagram, engineering drawing

Description automatically generated

**Figure S****9. Predicted recovery of LV size and function in response to removed volume overloading.** Similar arrangement for panels as in Figure 2. Growth module activated at 50 s when the simulation was at initial steady state. On all panels, first vertical line shows when the growth module is activated. Second and third vertical line demonstrate the onset and ending of the applied volume overloading. Fourth and fifth vertical lines shows the onset and ending of the removed volume overloading.

Diagram, schematic

Description automatically generated

**Figure S****10. Simulated aortic stenosis without the baroreflex control of arterial pressure.** The panels are arranged similarly to those in Figure 2. The simulation was gradually perturbed from 300 s to 400 s (second and third vertical dashed lines) by increasing Raorta (top panel in the right-hand column) in equation by 500%. The baroreflex module was deactivated at 200 s (vertical red dashed line).

Diagram, schematic

Description automatically generated

**Figure S****11. Simulated aortic insufficiency without the baroreflex control of arterial pressure.** The panels are arranged similarly to those in Figure 2, except that aortic regurgitant volume is shown in place of aortic resistance in the right-hand column. The simulation shown in this figure was perturbed gradually (second and third vertical dashed lines) by increasing Gaorta in equation from 0 to 1e-3 ([mmHg s]-1 L) to induce a regurgitant volume of ~40 ml (Table 3). The baroreflex module was deactivated at 200 s (vertical red dashed line).

Diagram, schematic

Description automatically generated

**Figure S****12. Simulated mitral insufficiency without the baroreflex control of arterial pressure.** The panels are arranged similarly to those in Figure 2, except that mitral regurgitant volume is shown in place of aortic resistance in the right-hand column. The simulation shown in this figure was perturbed gradually (second and third vertical dashed lines) by increasing Gmitral in equation from 0 to 2e-3 ([mmHg s]-1 L) to induce a mitral regurgitant volume of ~60 ml (Table 3). The baroreflex module was deactivated at 200 s (vertical red dashed line).