IBEHS 4A03: Biomedical Control Systems Project Report

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As a future member of the engineering profession, the student is responsible for performing the required work in an honest manner, without plagiarism and cheating. Submitting this work with my name and student number is a statement and understanding that this work is my own and adheres to the Academic Integrity Policy of McMaster University and the Code of Conduct of the Professional Engineers of Ontario. Submitted by **Faaria Khan, khanf67, 400135435**

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Background

Description of the system [1]

Arrythmias are irregularities in the heart's rate or rhythm that are caused by complications with the electrical impulses that coordinate the heart's beats. This can result in the heart beating too fast (tachycardia), too slow (bradycardia), or with irregularities (atrial/ventricular fibrillations).

One remediation therapy for patients with, or at risk of, arrythmias is a cardiac pacemaker. The pacemaker is subcutaneously implanted into the patient's heart. The device continually monitors the patient's cardiac rhythm and upon noticing the onset of an arrhythmic condition such as bradycardia, sends electrical impulses in place of the normal impulses to correct the rhythm. Some pacemakers are continually sending impulses to pace the heart in place of the sinoatrial node, while others only react when irregularity occurs. There are two components of a pacemaker, the leads, and the pulse generator. The leads are connected to the heart (the location and number of leads varies depending on patient condition) and send electrical signals back to the pulse generator. The pulse generator is responsible for filtering the received signal, analyzing it, and if necessary, generating a series of electric impulses to pace the heart rhythm. The signal received and filtered by the pulse generator represents ventricular depolarization shown by the QRS complex, providing insight to the cardiac cycle. The signal is then analyzed based off a normal rhythm and if the heart rate and arrhythmic duration exceed normal bounds, an arrythmia is considered detected. From here, onboard capacitors that store up charge can send an impulse through the leads to the heart to correct the arrythmia. The number of impulses, duration, and energy of the discharge varies depending on the condition and what was programmed specifically for that patient. This device can be used in many life-saving patient scenarios and can be in-vivo for years depending on the battery consumption.

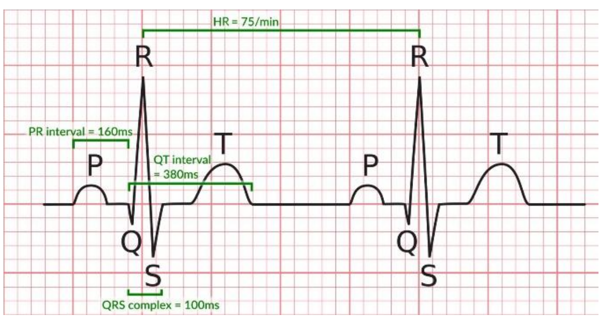


Figure 1. Sample QRS complex

A pacemaker contains all necessary components of a feedback control system and is what the following report is modeled after. Inputs to the system are the naturally occurring electrical impulses in the heart, detected by the leads. The leads relay the input to the pulse generator, which is the controller that processes the QRS complex, converts to a heart rate and assesses error in the measured input. The controller then outputs a corrective electrical impulse in response to the error to the actuator.

Review of the control strategy used in the source reference [2, 3]

In a source reference by Yadav et al., a cardiovascular system regulated by a cardiac pacemaker was modelled as a closed loop system with a controller and unity negative feedback.

The block diagram of the system consists of transfer functions for the controller $G_c(s)$, pacemaker element $G_p(s)$ and the heart $G_H(s)$.

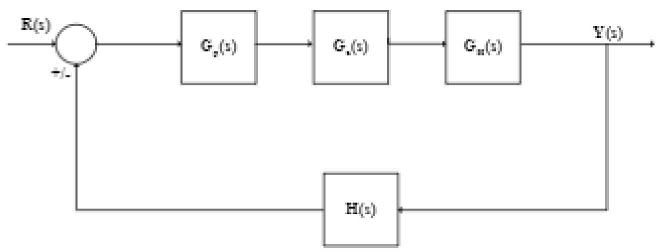


Figure 2. Source reference closed loop system with unity negative feedback.

The control strategy used by the pacemaker controller was that of a Proportional-Integral-Derivative (PID) controller. A PID controller entails calculating and controlling three parameters, the proportional, integral and derivative of the extent to which a process variable deviates from a desired set point, and in doing prescribes an input to an actuating element which minimizes deviations. The continuous-time PID controller prescribes an input $u(t)$ as follows:

$$u(t) = K_P\left(\epsilon(t) + \frac{1}{T_I}\int_0^t \epsilon(\tau)d\tau + K_D\frac{d\epsilon(t)}{dt}\right)$$

Where the Laplace transform is:

$$U(s) = K_P(E(s) + \frac{E(s)}{T_I s} + K_D sE(s)$$

The proportional component focuses only on the error between the set point and process variable and consists of a proportional gain K_C, which determines the ratio of the output response to the error signal and uses this to scale the system to increase the speed of control. The integral component sums the error over the duration of the system where it will continually increase unless the error is zero. Therefore, the integral component brings back the process variable to the setpoint from small deviations due to the oscillations. The derivative component predicts the rate of change of the process variable by taking the derivative of the error and can cause the system to react more strongly or weakly to prevent increasing error from occurring.

While a PID system is the most common controller strategy utilized in industry, other controller strategies such as Linear Quadratic Regulator (LQR) or Model Predictive Control (MPC) which offer higher levels of sophistication can be used in pacemaker models.

<u>Overview of the results found by the reference [3]</u>

The results obtained by Yadav et al. for their closed loop PID system, using a Zeigler-Nichols tuning method for various set point heart rates is as follows. For a more in-depth analysis into the reproduction of these results, please refer to the Results/Discussion section.

Table 4: Response parameter of PID controller tuned with

HR	Rise Time	Settling Time	Maximum % Overshoot
65	0.0714	1.0204	23.5488
75	0.0714	1.0195	23.4504
85	0.0711	1.0198	24.0782

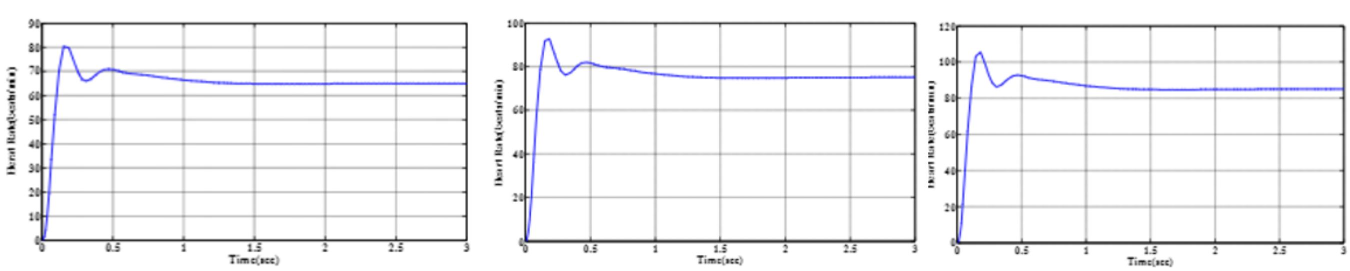


Figure 3. Response of the source reference PID controller tuned with Zeigler-Nichols for (1) HR = 65 BPM (2) HR = 75 BPM (3) HR = 85 BPM.

Simplifications/Assumptions

Pacemakers use electric impulses to keep a patient's heart beating at the correct rhythm and are typically only useful to treat persistent or sporadic bradycardia, which is a heart rate below 60 beats per minute. For cases where the patient has symptoms of tachycardia, a heart rate greater than 100 beats per minute, a device like an implantable cardiac defibrillator (ICD) would be more useful – where a defibrillating shock resets the SA node. To emulate this, our simulations will only encompass scenarios where the heart rate must be raised.

Another consideration is the allowable range that a patient's heart rate may take. The average person's resting heart rate is around 75 bpm and deviations too far above and below this for prolonged periods of time have serious health implications and may result in adverse events such as death. The purpose of the pacemaker we are simulating is to return to the set point heart rate as quickly as possible when a disturbance such as a bradycardic episode is introduced. However, we must be aware of the overshoot and undershoot caused by the PIDs error correction and make sure the heart rate does not exceed or fall below 90 and 60 BPM. In addition, the rapid changes in heart rate must occur in an appropriate time span – otherwise this may manifest symptoms of atrial flutter such as tightness, shortness of breath or dizziness. The above considerations can be met by efficient tuning of the PID controller.

Methodology

Mathematical model and description

The base mathematical model of our pacemaker control system was derived from Yadav et al. and follows the below block diagram,

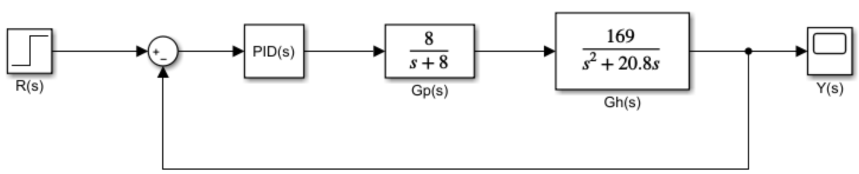


Figure 4. Closed loop Simulink model of pacemaker system.

The input to the system, R(s), represents the set point heart rate that the system aims to return to and stay at after a deviation. The output is Y(s), which is the patient's true outputted heart rate measured by leads in the heart. Within the system loop are three transfer functions:

1) $G_C(s)$ is the PID controller, whose parameters are determined using tuning and optimization methods.
2) $G_P(s)$ is the pacemaker actuating element with system dynamics: $\frac{8}{s+8}$
   The transfer function for the pacemaker is a low pass filter which serves to filter out higher frequency signals while allowing lower frequency ones.
3) $G_H(s)$ is the transfer function for the heart and is represented as an under damped second order system that mimics the heart's normal function: $\frac{169}{s^2+20.8s}$

To make the base mathematical model of the pacemaker more realistic, we have introduced further system dynamics.

AV Delay [4]:
In the case of normal conduction within the heart, the sinoatrial (SA) node, otherwise known as the heart's *natural* pacemaker, sends out an electrical impulse for the atria to contract and triggers the atrioventricular node (AV). Subsequently, the AV node sends out an impulse causing the ventricles to contract. In an ECG, these events can be seen through a PQRST complex. Specifically, the PR interval is of interest to pacemaker manufacturers, as it is the period that extends from the beginning of atrial depolarization until the beginning of the onset of ventricular depolarization. In dual chamber pacemaking modes, there must be a programmed atrioventricular (AV) delay between an atrial event (sensed or paced) and a scheduled paced ventricular event – equivalent to a PR interval under healthy circumstances. This AV delay must be optimized and carefully programmed to ensure the atria has enough time to empty blood into the ventricles before the ventricles contract.

In our updated model, we introduced an AV delay to maintain mechanical synchronization of the cardiac chambers. A study by Statescu, et al. compared the nominally programmed AV delay of 125-175 milliseconds to optimized delay intervals for a sample size of 55 patients, with a mean age of 67, all with initial dual chamber pacemakers implanted. Through various tests evaluating ventricle end diastolic and systolic volumes, ejection fractions, stroke volumes, Doppler echocardiography of the aortic and mitral valves, it was determined that the optimal AV delay achieved on the programming of the pacemaker was 145-250 milliseconds [2]. We implemented a 145-millisecond delay as a process delay, with a lag between the control input signal to the Gp(s) block to delay the controller in actuating an electrical impulse.

Electromagnetic Interference [5]:
Increasing technological advances have resulted in new sources of electromagnetic radiation, which may pose a threat to implantable pacemakers. Generally, EMI results in pacing inhibition, despite pacemakers using bipolar sensing and low pass filters to reduce conducted and radiated interference [3]. To mimic this in our model, we implemented a dampening factor in the controller output which will cause the current proportional gain according to the calculated error to be understated.

Thus, we arrive at the following pacemaker model:

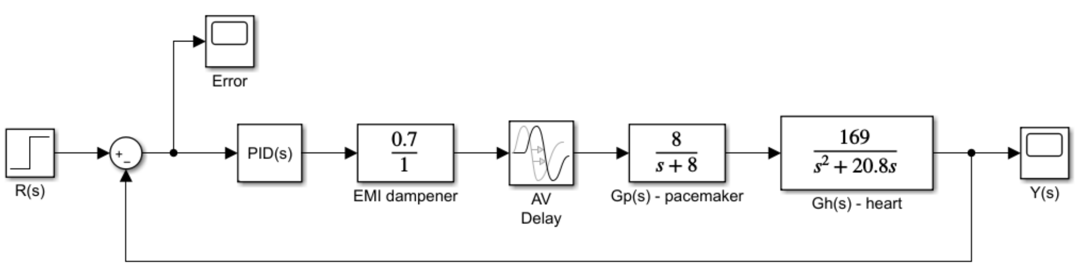


Figure 5. Updated pacemaker Simulink model.

<u>Parameters selected for deviation and how they relate to real system</u>
To test our model's efficacy as an actual pacemaker, we introduced a deviation to its set point. This comes in the form of a bradycardic disturbance to a resting heart rate. Bradycardia is when the heart rate drops below 60 bpm and can have health implications as the body is not receiving adequate oxygenated blood. In our model, we designed this disturbance as a step function that goes from the regular 75 bpm heart rate to 40 bpm after a certain step time. The block for this disturbance was added as an input into the loop right before the outputted heart rate. This way it alters the output heart rate, and the feedback registers the deviation as an error that must be

corrected. In a real pacemaker, the heart rate is continually being monitored and when a bradycardic arrythmia occurs it is sensed, and the pacemaker responds immediately by using electric impulses to raise the heart rate. Ideally, our model responds in the same way and corrects for the disturbance quickly without much oscillation. We found that there is an immediate drop to 40 bpm when the deviation is first applied and when the output is compared to the set point heart rate, the PID promptly attempts to return to the set point. Before tuning the PID, the model still recovers from the bradycardic disturbance, but takes too long and there is problematic overshoot/undershoot.

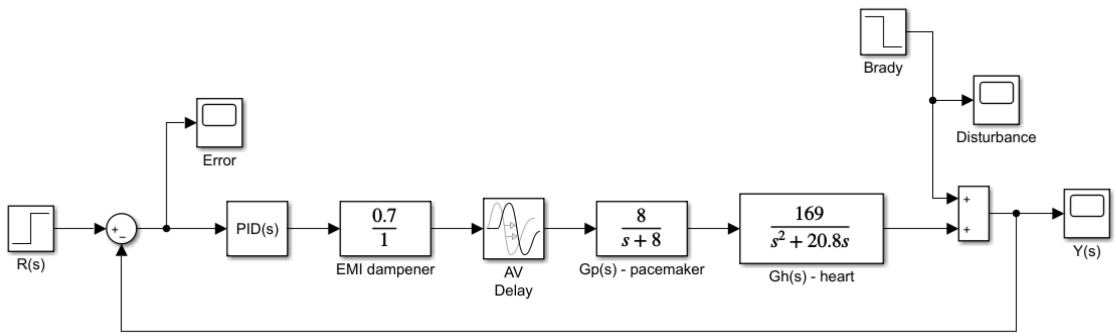


Figure 6. Pacemaker Simulink Model with Disturbance.

<u>Discussion of implementation in MATLAB/Simulink</u>

Our model of the pacemaker was created and simulated in Simulink using its block diagram functionality. Blocks for each transfer function and component were aligned as either input, outputs, or parts of the loop and by running the simulation for a set amount of time we could see the scopes of the output and inputs. MATLAB was used adjacently with the Simulink model to either calculate values we would use when designing the block model, or to receive the simulations in the script workspace to analyze them further.

With our selected model for the pacemaker, we began to look to optimize the output of the closed loop system. This is done by tuning the PID controller of the closed loop system so that it minimizes the error. Tuning is the process by which we take the K_P, K_I, and K_D parameters for our PID controller and manipulate their values until we find the combination of values that optimizes the performance. In this case, the optimized performance for a pacemaker would be one in which the heart rate is promptly brought back to a normal heart rate of 75 bpm, with minimal overshoot/undershoot. To do this however, we need initial guesses of their values that serve as a starting point for the tuning process. We chose to use the Ziegler-Nichols tuning procedure, which utilizes a frequency stability analysis and predetermined operations applied to the values obtained to give a solid starting guess.

The first step in the Ziegler-Nichols procedure is to create a Bode diagram from the transfer function of the feedback loop in our model.

$$G_{loop} = \frac{946.4e^{-0.145s}}{(s+8)(s^2+20.8s)}$$

In our calculation for the loop transfer function, the controller was assumed to be proportional-only with K_C=1. By using a script in MATLAB (refer to Appendix), we inputted this transfer function and obtained the following Bode plot. It is split into two graphs, where the top one compares the functions gain to its input frequency, and the bottom one compares the phase shift that occurs when changing input frequency.

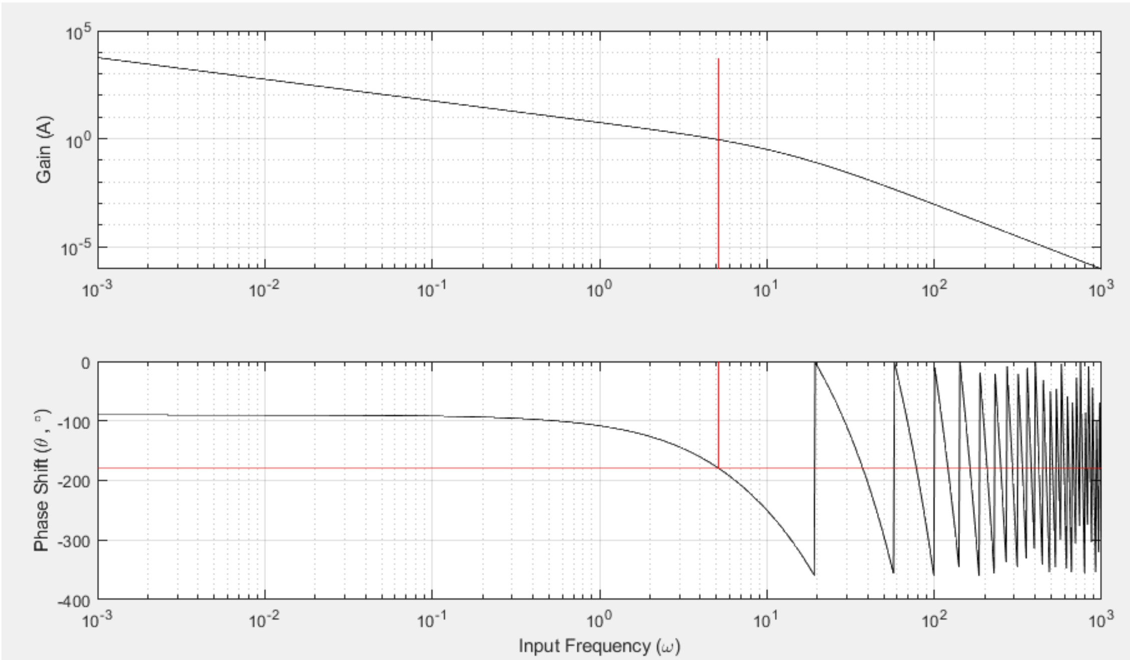


Figure 7. Bode stability analysis of the system used for Zeigler-Nichols tuning.

Using the Bode diagram, we can now find the critical frequency and amplitude at the critical frequency. The critical frequency of the system is found by looking for point at which there is a phase shift of -180 degrees. Then by evaluating the loop transfer function at $j\omega_C$ and afterwards finding the norm of that function, we obtain the critical amplitude.

The critical frequency of the system that we found was 5.1489 radians.

$$\omega_C = 5.1489\, rad$$

First, we substitute $j\omega_C$ for s in the loop function.

$$G_{loop}(j\omega_C) = \frac{946.4e^{-0.145j\omega_C}}{(j\omega_C + 8)(j\omega_C{}^2 + 20.8j\omega_C)}$$

Next, we find the critical amplitude by taking the norm of the loop transfer function.

$$A_C = |\, G_{loop}(j\omega_C)\,| * K_C\ , where\ K_C = 1$$
$$= 0.9016$$

Now with the critical values, we can calculate the gain and period of the gain that would cause marginal stability, the ultimate gain and ultimate period.

$$K_U = \frac{1}{|\, G_{loop}(j\omega_C)\,|} = \frac{1}{0.9016}$$

$$= 1.1091$$

$$T_U = \frac{2\pi}{\omega_C} = \frac{2\pi}{5.1489}$$
$$= 1.2203$$

These values are then used to find the initial guesses for the closed-loop tuning parameters of the PID controller by applying predetermined coefficients.

$$K_C = \frac{K_U}{1.7} = \frac{1.1091}{1.7}$$
$$= 0.6524$$

$$\tau_I = \frac{T_U}{2.0} = \frac{1.2203}{2.0}$$
$$= 0.6102$$

$$K_D = \frac{T_U}{8.0} = \frac{1.2203}{8.0}$$
$$= 0.1525$$

Finally, we get the PID controller transfer function tuned using the Ziegler-Nichols method, derived from a Bode stability analysis.

$$G_C = 0.6524(1 + \frac{1}{0.6102s} + 0.1525s)$$

Now using Simulink, we can update the parameters of the PID controller in our model and compare the output to the previously untuned system.

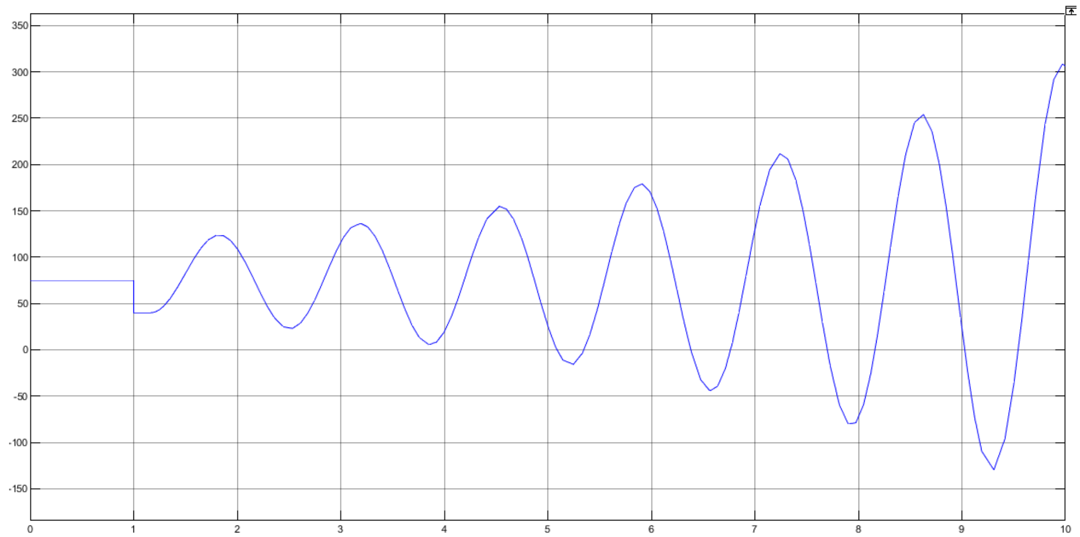


Figure 8. Simulink scope plot of the output heart rate vs. time (seconds) before initial tuning.

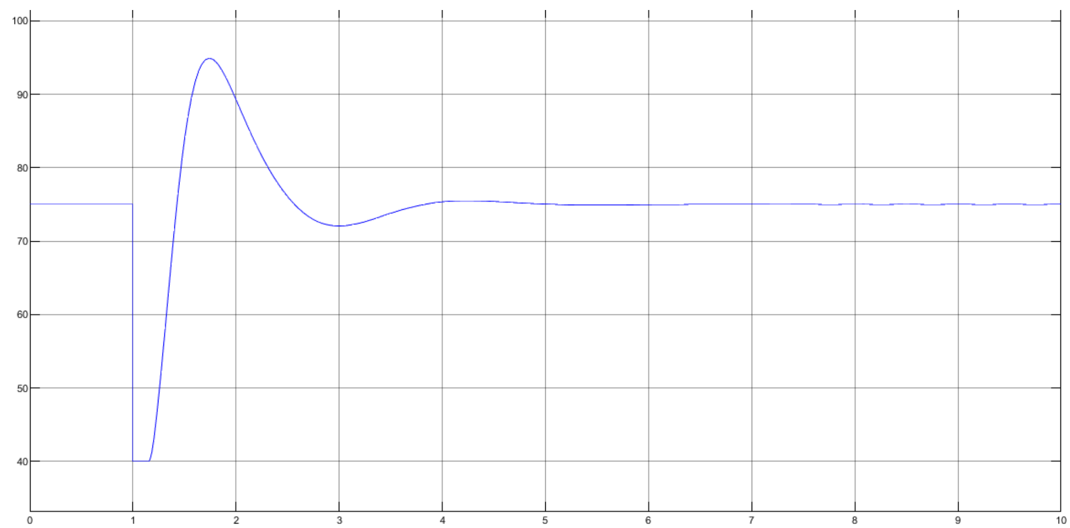


Figure 9. Simulink scope plot of the output heart rate vs. time (seconds) after initial tuning.

In the first simulation, we have the model of the pacemaker with the default parameters for the PID that Simulink provides (K_P=1, K_I=1, and K_D=0). The output Y(s) shows that with these controller parameters the signal is unstable, and the heart rate never reaches the desired set point. Next, we showed the output heart rate for the Ziegler-Nichols tuned system. Immediately we see an improvement as the heart rate returns to the 75-bpm set point after the bradycardic disturbance as opposed to the untuned system. This is promising however the parameters need to be further tuned still. We see that after the disturbance, the controller attempts to correct for the drop and greatly overshoots the intended set point. Once we optimize the tuning parameters, the PID should have its overshoot and oscillations minimized. The parameters also need be adjusted further to reach the set point faster like a proper pacemaker would.

<u>Optimization using IAE</u>

Integral absolute error (IAE) integrates the absolute error over time, where the error is defined as the difference between the actual output, and the desired output. No weight is added to any specific error, so all errors are penalized equally. To optimize the PID controls, multiple simulations using variable controller values were run on the system, with smaller IAE values being considered better. Using the results from the Ziegler Nichols tuning as a reference, the proportional gain varied from 0.4 to 0.8, with 0.01 intervals. The integral gain, after initial testing, caused too much overshoot for our system with any value greater than 0.4, so values between 0 to 0.4, at 0.01 intervals were chosen. The derivative gain was at a constant 0.1525 for all iterations, to conserve simulation run time.

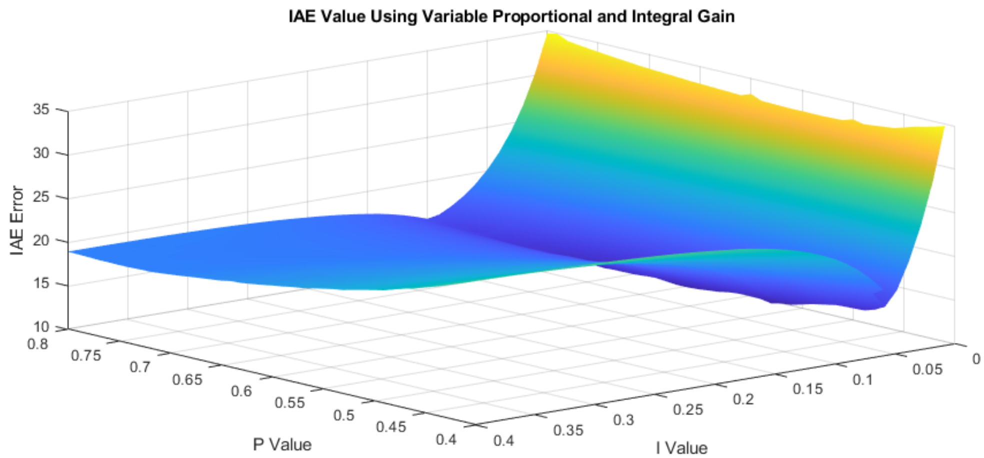


Figure 10: Surface plot of the Integral Absolute Error (IAE) when variable P and I values are used. D is left constant at 0.1525. Ideal IAE values are visually identifiable in the dips of the surface, at around I= 0.06. The minimum IAE value is 13.00493, occurring with PID values corresponding to (P=0.51, I=0.06, D=0.1525).

Using the PID values associated with the minimum IAE, a significant improvement in overshoot and time needed to reach a steady state heart rate of 75 BPM is seen in Figure 11. These optimized parameters ensure safe operation of the electrical activity within the heart.

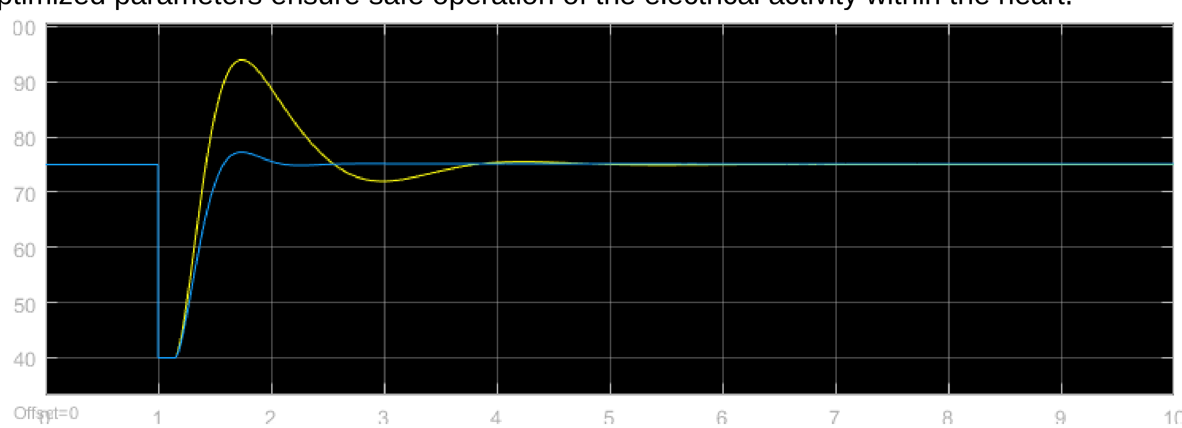


Figure 11: Output heart rate using the optimized PID values (blue), found through minimizing the IAE, compared to Ziegler-Nichols tuning (yellow).

Feedforward Variation

One proposed variation of the model that may enhance the pacemaker performance is a PID system with feedforward control. Traditional PID controllers can only compensatory input to an actuating element when the process variable has deviated relative to the set point, producing a controller error: $\epsilon(t) = set\ point - process\ variable$. However, this means deviation of the process variable must propagate throughout the entire system to be recognized by the feedback controller and the controller action at best only works to remedy the negative effects of the disruption to the system. In contrast, a feedforward controller measures a disturbance directly and predicts preemptive control actions before an impact is made to the process variable.

In our pacemaker system, we try to anticipate the impact of a bradycardic heart rate on the patient before it occurs. To obtain the feedforward transfer function, we use the following block diagram algebra:

$$Y(s) = (Brady)(e^{-0.145s}) + (Brady)(G_{FF})(0.7)(e^{-0.145s})\left(\frac{8}{s+8}\right)\left(\frac{169}{s^2+20.8s}\right)$$

$$0 = (Brady)(e^{-0.145s})\left[1 + (G_{FF})(0.7)\left(\frac{8}{s+8}\right)\left(\frac{169}{s^2+20.8s}\right)\right]$$

$$G_{FF} = -\frac{1}{(0.7)\left(\frac{8}{s+8}\right)\left(\frac{169}{s^2+20.8s}\right)} \rightarrow simplifed\ to\ G_{FF} = \frac{-\frac{1}{946.4}}{s^3 + 28.8s^2 + 166.4s}$$

It should be noted that feedforward control is not physically realizable when there is a sensor delay but no delay of the disturbance on the process. Hence, additional circuitry to introduce a delay in the bradycardic disturbance must be installed, yielding a feedforward system such as,

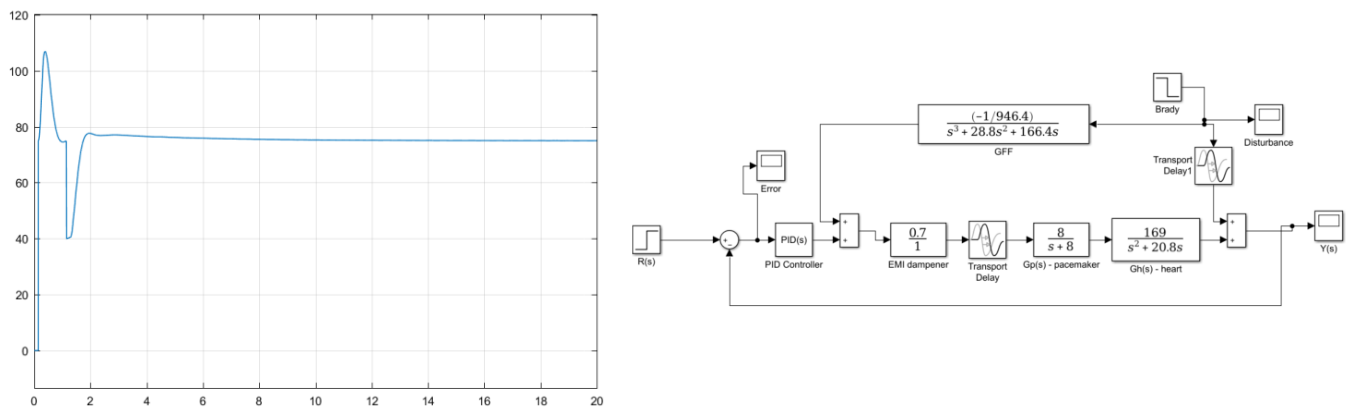


Figure 12. Variation of the pacemaker control system with a feedforward response to the disturbance.

A feedforward controller in our pacemaker model is not suitable, as seen in Figure 12. The feedforward loop anticipates the impact of the bradycardia rhythm on the heart rate process variable and tries to counteract by overshooting to a higher heart rate prior to the onset of the bradycardic rhythm. As discussed before, the average person's resting heart rate is around 75 BPM and overshoot/undershoot should not exceed or fall below 90 and 60 BPM as this may manifest itself into health implications for the patient. Another limitation that deems feedforward control unsuitable for this pacemaker model is the fact that feedforward control can only be used for a very specific and known bradycardic disturbance, meaning feedforward control would not be able to offer any mitigation to patients with unpredictable bradycardic rhythms.

Results and Discussion

Reproduction of literature results

While the source reference by Yadav et al. model of the cardiovascular system regulated by a cardiac pacemaker was less complex than our own, their focus was to demonstrate the efficacy of different tuning methods. Here we reproduced their system with the Ziegler-Nichols method that they outlined, which varied from how we approached Ziegler-Nichols tuning.

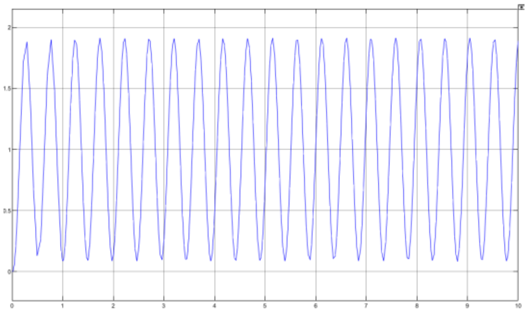


Figure 13. Critical response of Y(s)

Type of controller	K_c	T_i	T_d
P	$0.5K_{cr}$	∞	0
PI	$0.45K_{cr}$	$\frac{1}{1.2}P_{cr}$	0
PID	$0.6K_{cr}$	$0.5P_{cr}$	$0.125P_{cr}$

Figure 14. Zeigler Nichols parameter rules.

To find the K_P, K_I, and K_D parameters for the PID we followed the following Ziegler-Nichols steps the paper outlined. First, taking the transfer function of the PID, $C(s) = K_P + \frac{K_I}{s} + K_D S = K_P(1 + \frac{1}{T_I S} + T_D S)$, we set the values of T_I to ∞ and T_D to 0. Now that the function is just equal to the gain K_P, we started from K_P=0 and kept increasing its gain until the point that the system was about to be unstable. This value is called the critical gain, K_{CR}. From here, the period between oscillations was determined, P_{CR}, and the parameters can be found with the following table. Using this method, we determined that for their model K_{CR} = 3.545 and P_{CR} = 0.48sec. With this we found the tuned parameters to be K_P = 0.288, K_I = 0.24, and K_D = 0.06.

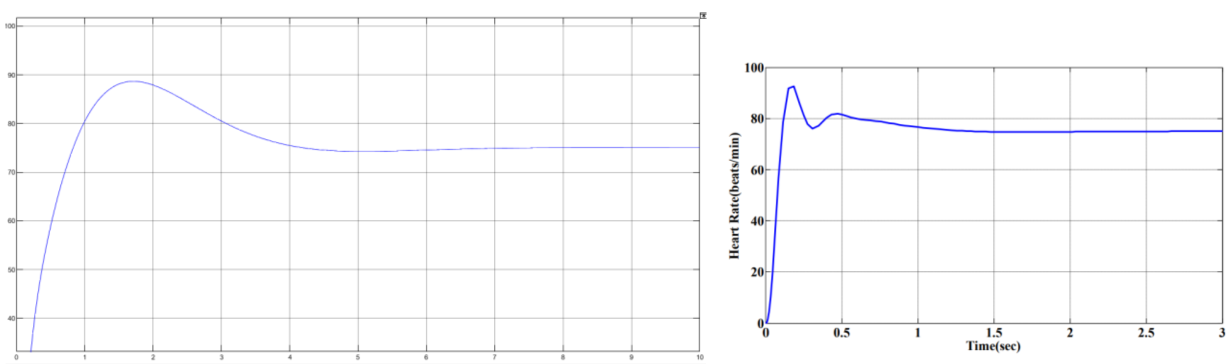
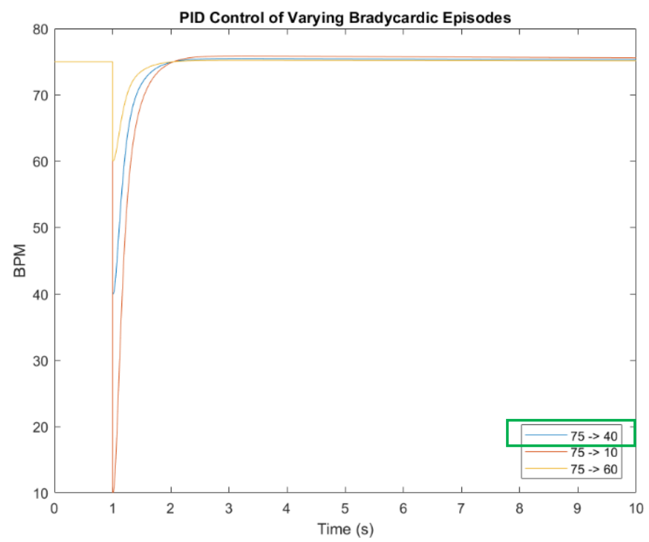


Figure 15. Scope of Y(s) with ZN tuned controller; our reproduction (left) compared to paper (right).

Above is the plot of the output heart rate from the system with the paper's Ziegler-Nichols tuned parameters. It does not exactly match with the plot included in the paper, where both our model and theirs had a set point of 75 bpm. This is likely because we emulated the steps that they outlined to follow for tuning, but their method was unprecise infringing on its reproducibility. We were able to confirm that the model does in fact work, and it helped as a good reference when we expanded the design to better model a real pacemaker.

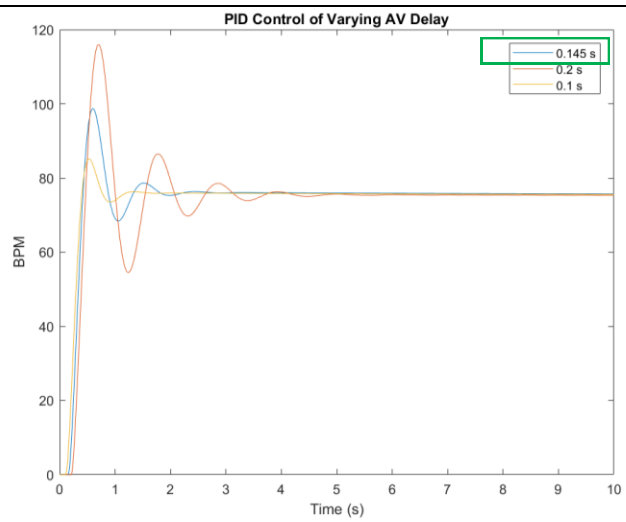
Base Case Results with Fixed Controller Parameters and Variable Modeling Parameters

Case 1: PID Control of Bradycardic Disturbance ONLY



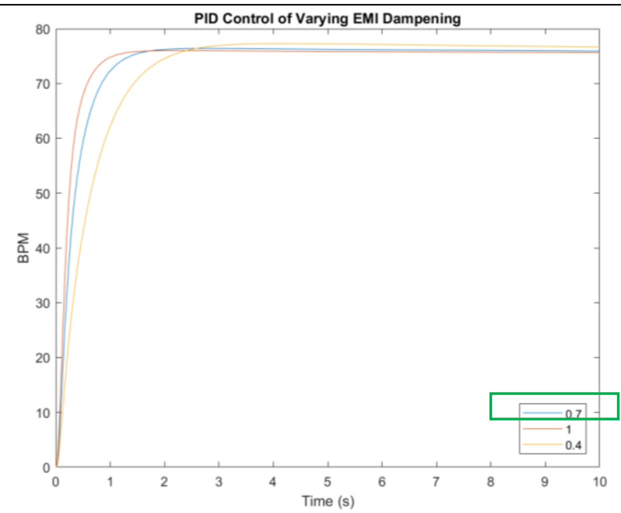
To demonstrate our pacemaker model's robustness, we introduced a bradycardic disturbance rhythm in the form of a step function that goes from 75 to 40 BPM at time step 1. Our optimized PID model can handle a variety of bradycardic rhythms, such as 60, 40 or 10 BPM to reflect a variety of patient cases. In all cases, the system response is critically damped, however disturbances with smaller deviations from the setpoint heart rate of 75 BPM have faster settling times due to the proportional gain scaling the system response relative to the error.

Case 2: PID Control of AV Delay ONLY



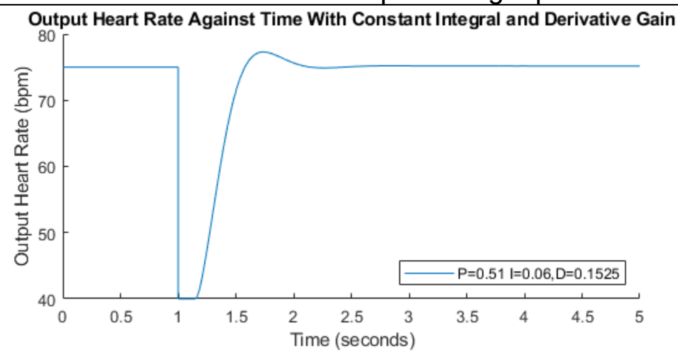
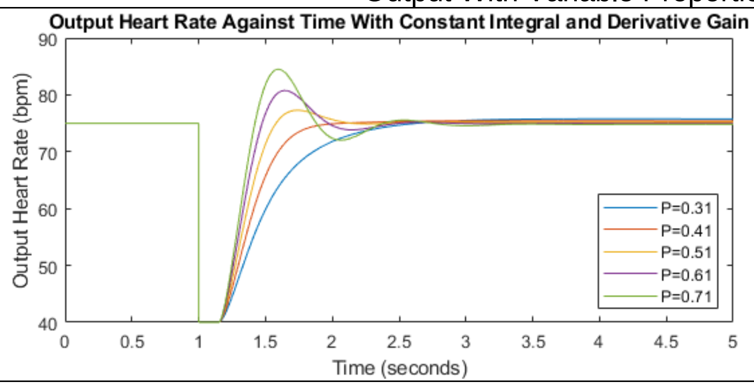
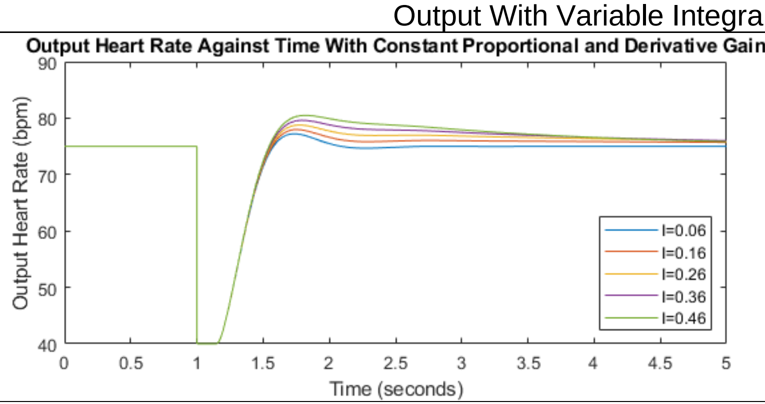
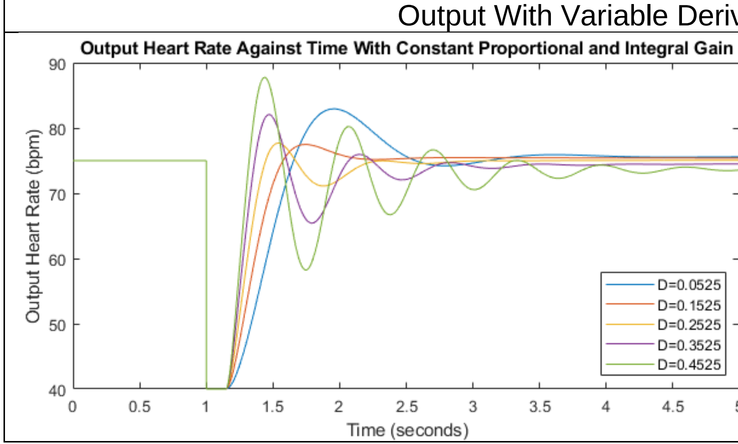
The atrioventricular (AV) delay is implemented as a process delay between the control input signal to the actuator, to ensure the atria has enough time to empty before a subsequent electrical impulse is fired. The trend demonstrated in increasing the AV delay is imposing larger oscillations. This is a result of the calculated error between the setpoint heart rate and delayed feedback heart rate not accurately reflecting the true error between the setpoint and current output heart rate, and overestimating/underestimating K_P.

Case 3: PID Control of EMI ONLY



Exemplified are 3 degrees of electromagnetic interference, which have an inhibitory effect on the pacemaker actuation. A dampening constant of 0.7 represents someone who typically only comes across household electronics, whereas 0.4 represents someone who works in an occupation with aircrafts, high voltage power coils or magnets. A dampening constant of 1, represents no EMI and is unlikely. The trend of increasing EMI dampening results in understated proportional gain which slows the system response.

Case Study and Controller Manipulation

Output Using Optimized PID Controller	
	Shown for reference is the optimized system with the modeling parameters enclosed in green boxes above, using values: 1. P = 0.51 2. I = 0.06 3. D = 0.1525 When altering PID values below, variables that are kept constant will use these listed values.
Output With Variable Proportional Gain	
	With all else constant aside from the proportional gain (P), increasing P decreases the rise time. The resulting overshoot increases, and the amount of time it takes to settle slightly increases.
Output With Variable Integral Gain	
	By increasing the integral gain, a slight increase in overshoot is observed. There is also an increase in the amount of time required for the output to settle to its steady state.
Output With Variable Derivative Gain	
	Unlike the other controller parameters, changing D in this scenario shows that there is a sweet spot. By D, the overshoot significantly increases, and displays an underdamped response. By decreasing D, this oscillation disappears, but at the cost of a slower rise time.

References

[1] M. Glikson and P. A. Friedman, “The implantable cardioverter defibrillator,” The Lancet, vol. 357, no. 9262, pp. 1107–1117, 2001.

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[3] J. Yadav, A. Rani, and G. Garg, “Intelligent Heart Rate Controller for Cardiac Pacemaker,” International Journal of Computer Applications (0975 – 8887), vol. 36, no. 7, Nov. 2013.

[4] Statescu C, Sascau RA, Maciuc V, Arsenescu Georgescu C. Programming an optimal atrioventricular interval in a dual chamber pacemaker regional population. Maedica (Bucur). 2011; 6(4): 272-276.

[5] Beinart R, Nazarian S. Effects of external electrical and magnetic fields on pacemakers and defibrillators: from engineering principles to clinical practice. Circulation. 2013;128(25):2799-2809. doi:10.1161/CIRCULATIONAHA.113.005697