LOW FREQUENCY BIMORPH CANTILEVER ENERGY HARVESTER FOR PACEMAKER APPLICATIONS

by

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Declaration of Authorship

I declare that this thesis is my own work and, to the best of my knowledge and belief, it does not contain material published or written by a third party, except where permission has been obtained and/or appropriately cited through full and accurate referencing.

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Date…..30th of May, 2020…..

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Dedication

To my family…

Abstract

Pacemakers have seen multiple technological advancements but still face multiple issues. The ability to have a leadless and self-powered pacemaker has been always sought after by researchers and scientists. The introduction of flexible electronics has made such an endeavour a realistic and achievable one since it allows pacemakers implantation to take another advantageous shape that allows it to be further useful in the process of implantation without injuring surrounding tissue. Furthermore, energy harvesting systems have developed to a point where a pacemaker can obtain its power from naturally occurring motions and other phenomena inside the body. In this research several techniques and frameworks have been studied to realize a leadless, flexible, self-powered and smart single chamber pacemaker. The main objective of this thesis is to provide a model of a device that consumes minimum amount of power, enabling energy harvesting to be applied. This hence improves the quality of life of patients and deliver acceptable performance as compared to previously made pacemakers.

Keywords: ***Flexible Electronics; pacemaker; Energy Harvesting.***

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List of Abbreviations

E_p	Young's modulus of piezoelectric material
E_s	Young's modulus of substrate
l_m	Length of proof mass
m_p	Proof mass
s^E	Elastic compliance [Pa^{-1}]
t_p	Thickness of piezoelectric material
t_s	Thickness of substrate
ϵ^T	Dielectric permittivity [F/m]
A	Amplitude of signal
AV Node	Atrioventricular Node
BPM	Beats Per Minute
ECG	Electrocardiogram
K	Spring constant
LA	Left Atrium
LV	Left Ventricle
m	Effective mass of the cantilever
mc	Cantilever mass
mt	Tip mass
OPAMP	Operational Amplifier
PCB	Printed Circuit Board
Pk-pk	Peak to peak
Q	Quality factor

RA	Right Atrium
RV	Right Ventricle
SA Node	Sinoatrial Node
t	Thickness of cantilever
W	Width of cantilever
Wn	Resonance frequency
Y	Young's modulus
D	Electric displacement [C/m^2]
E	Electric field [V/m]
S	Mechanical Strain
T	Mechanical stress [N/m^2]
d	Piezoelectric coefficient [C/N]
l	Length of the beam
s	Density of substrate material
w	Width of the beam
ρ	Density of piezoelectric material

Chapter 1. Introduction

In this chapter, an introduction about the current standards of pacemakers and their mechanisms as well as energy harvesters and the physics behind their operation is given. Heart anatomy and physiology is also explained in detail. Then, we present the problems investigated with current pacemakers and our thesis contribution. Finally, general organization of the thesis is presented.

1.1. Overview

Advances in electronics and energy harvesting systems have led to a huge interest in developing a truly leadless and self-powered pacemaker. However, this interest was not realized due to the many obstacles facing it, mainly the reliability of the input power needed inside the pacemaker in order to function, as well as the method of implantation. These reasons have led to the limitation in using such pacemakers. This has pushed researchers into developing new devices and energy harvesters that would utilize the mechanical energy within the heart (contractions) in order to improve the reliability of pacemakers that would rely on it. Another reason leadless is sought after with such interest is that it would potentially remove the issues with lead and pockets that exist in current pacemakers [1].

1.2. Thesis Objectives

Driven by the current advancements in technology and the desire to improve patients' quality of life, we will focus on the development of a low frequency bimorph cantilever energy harvester for pacemaker applications. The material for the device will be chosen to be flexible enough to contract with the heart. Furthermore, the energy harvesting system parameters will be chosen to utilize the energy used by the heart. The main objective of this work is to investigate the design of a complete energy harvesting device that adheres to the current standards of power needs for pacemakers.

1.3. Research Contribution

The contributions of this research work can be summarized as follows:

- Propose an energy harvesting system that will use heart-energy to power the device.
- Propose an energy storage circuit to help the pacemaker operate more reliably.

- Propose a fabrication process for the device.
- Indicate the method of insertion for the device anatomically.

1.4. Thesis Organization

The thesis later on is organized as follows: Chapter 2 provides background about the heart anatomy and physiology, as well as the current pacemaker technology and energy harvester systems. The device simulation with all its individual parts are discussed in Chapter 3 along with the proposed schematic. Moreover, Chapter 4 will contain the experimental stage, where all the testing and testing conditions will be mentioned. Finally, the results and analysis section will be in Chapter 5 which will conclude this thesis and outlines future work.

Chapter 2. Background and Literature Review

In this chapter, we discuss the anatomy and physiology of the heart as well as the current technology of pacemakers along with its architecture. We will then present the techniques that are used as well as the modes it operates on. Energy harvesting devices will also be discussed with regards to their use in modern pacemakers. Furthermore, the standards will be mentioned as a metric to evaluate our device performance. Finally, all related works in the literature will be discussed. However, since this subject is still under research and there is no well-known standard established for this type of device, the amount of references that exist in this specific topic are very rare.

2.1. Heart Anatomy

The human heart is a marvellous pumping mechanism that work 24 hours, 7 days a week nonstop. It is one of the most important organs in our body due to its crucial function, which is to pump and deliver blood to every vascularized part of our body. We need efficient blood delivery to every part of our body because without blood our tissues would not have access to oxygen, and they would quickly build up carbon dioxide and die. The heart anatomy is quite simple, as shown in Figure 1, it consists of 4 chambers, the atria (2 upper chambers) and the ventricles (2 lower chambers).

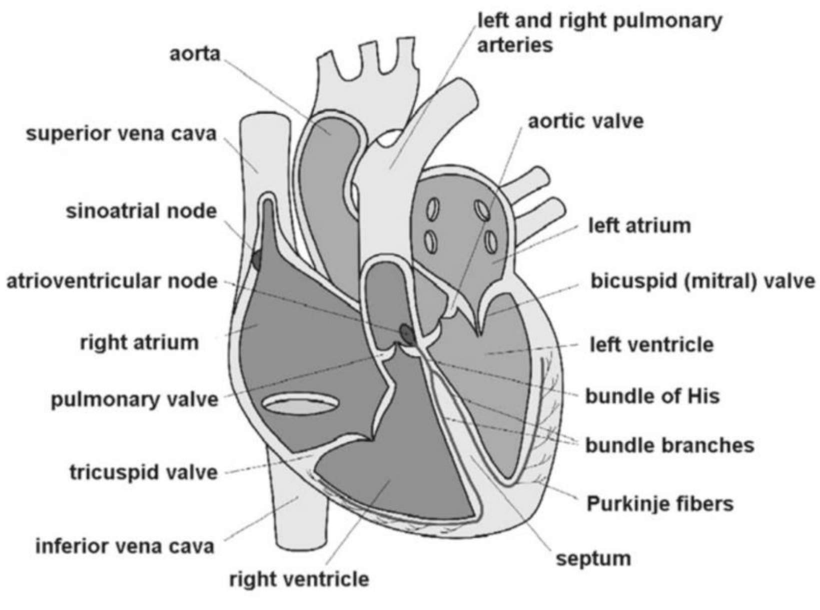

Figure 1: Heart Anatomical Structure [2]

The right and the left side of the heart are separated by a wall (the septum), valves exist between the atria and ventricles (atrioventricular [AV] valves) and between the ventricles and the pulmonary artery and vein (semilunar valves), we can clearly see that as shown in Figure 2.

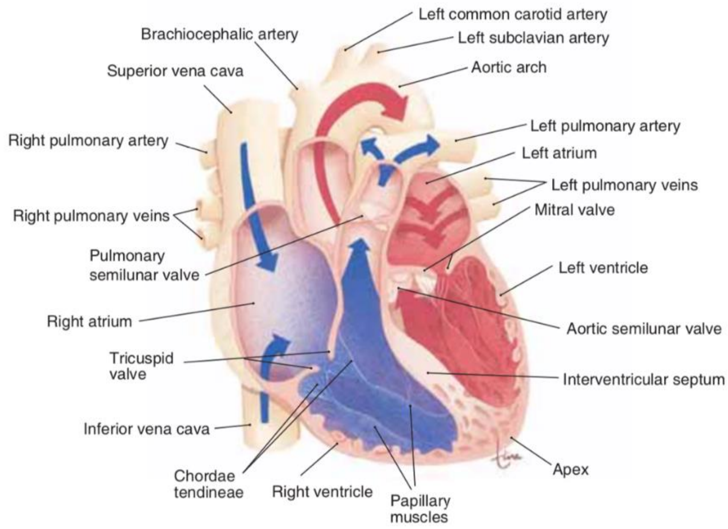

Figure 2: Blood cycle inside the heart chambers [3]

The cardiac cycle starts when the right atrium fills up with deoxygenated blood from the veins, pressure builds up and the atrium contracts to push the blood to the right ventricle. The right ventricle fills up then pushes the blood through the pulmonary artery to pass it through the lungs and oxygenate the blood. The blood passes the pulmonary circuit and comes back to the heart through the pulmonary vein to the left atrium. The left atrium pushes the blood to the left ventricle where it then pushes the oxygenated blood to the aorta and to the rest of the body [2]. This is illustrated in Figure 3.

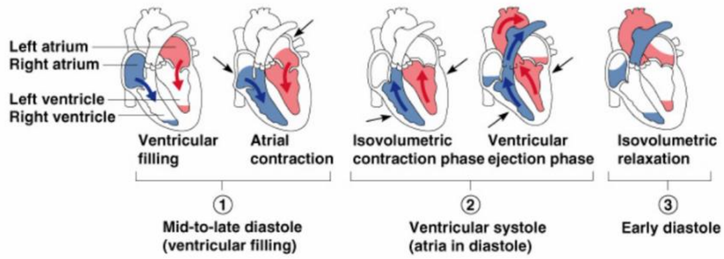

Figure 3: Cardiac contraction cycle [3]

The heart also beats on average 60-100 times per minute, this is the BPM or beats per minute. The coordination done between all the different parts of the heart is through its internal electrical system. The heart paces itself through electrical signal propagation from one node to the next.

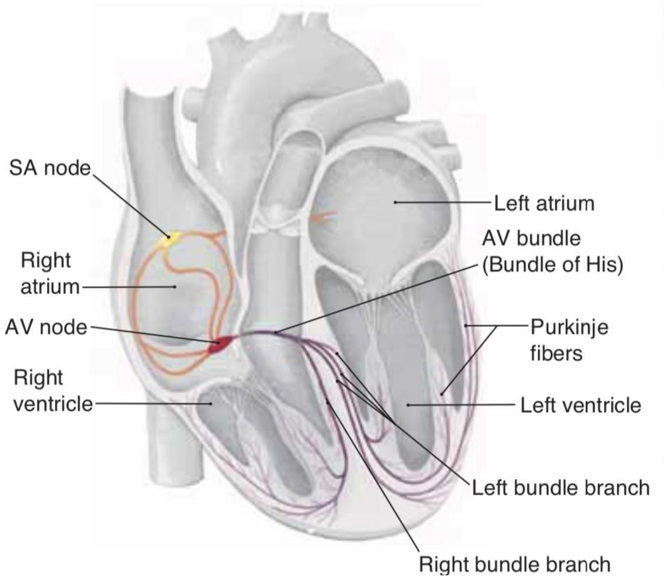

Figure 4: The cardiac electric system [3]

The signal starts from the sinoatrial (SA) node, as shown in Figure 4, which is also called the pacemaker of the heart, and then into the atrioventricular node, to the atrioventricular bundle and bundle branches into the Purkinje fibres.

Such an intricate and complex system may malfunction, and this causes numerous diseases and syndromes. Cardiovascular diseases are the cause of 1 of every 3 deaths in the US [3].

2.1.1. Heart diseases. Some of the diseases that can occur in the heart belong to arrhythmias, which is characterized by having abnormal heart rhythm. Tachycardia is when the heart is beating faster than average (>100 bpm), bradycardia is when the heart is beating slower than usual (<60 bpm). Both tachycardia and bradycardia are forms of arrhythmias that occur in the heart. Arrhythmias can be very dangerous, causing fatigue, dizziness and can lead to death. We can see a further breakdown of heart diseases in Figure 5.

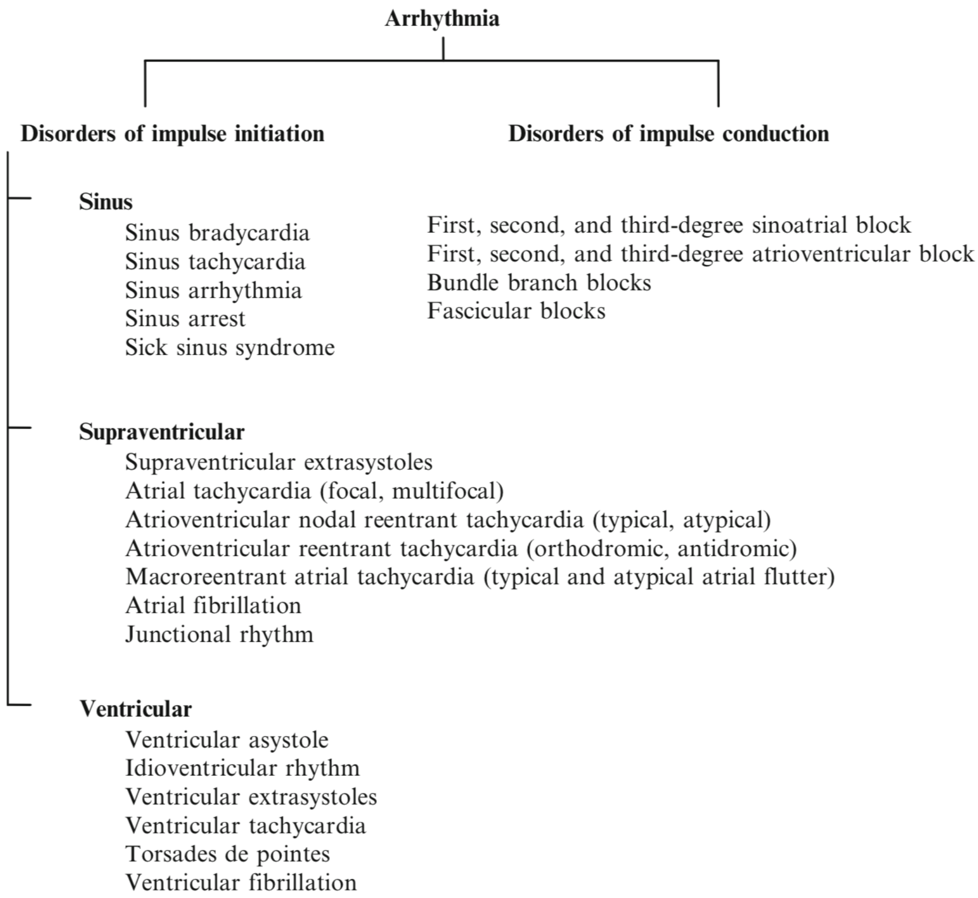

Figure 5: Heart Arrhythmia types [2]

These diseases might develop into a serious case where a cardiac pacemaker device need to be implanted for the human to be able to sustain life.

2.2. Muscle Cell Polarization

The cell body is enclosed by a double phospholipid layer that act as a cell boundary. Its main job is to enclose and maintain the intracellular environment. The extracellular environment contains electrolytes and other materials that is transported inside the cell if needed. Inside the cell, potassium ions are predominant with a concentration about 30 times higher than the extracellular matrix. The cell double layer is permeable to potassium ions. On the other hand, in the extracellular matrix sodium ions are higher than the intracellular matrix. What controls the cell membrane permeability is the presence of calcium ions inside the cell. The cell strives to reach and manage these ions in order to achieve thermodynamic balance. Since ions on either side of the cell exist an electric field is generated, at rest it is called the resting membrane potential. In Figure 6, we can see that the extracellular region in this case is positively charged, while it is negatively charged inside the cell. In the case of myocardial cells, the typical resting membrane potential is -90 mV, while in the SA node it is about -45 mV [4].

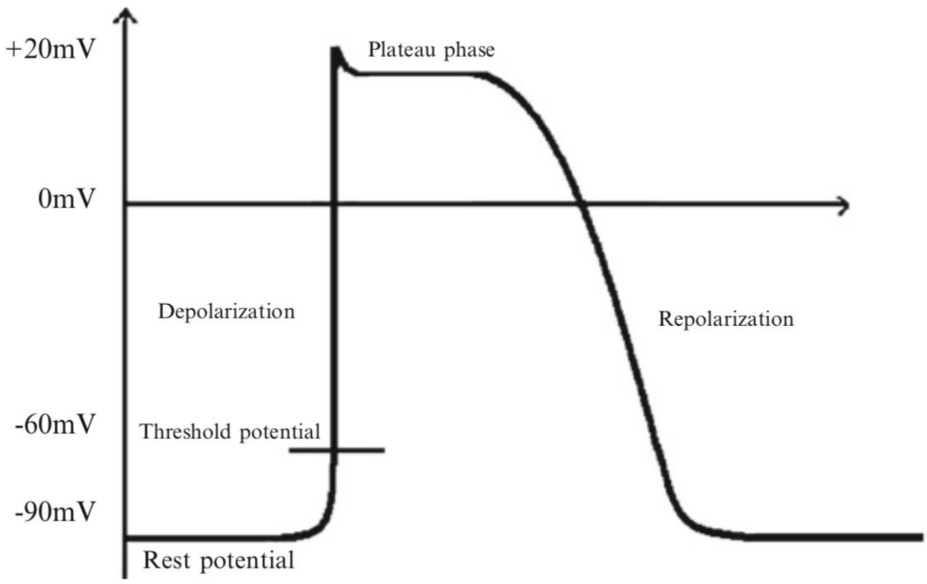

Figure 6: Action potential signal [2]

The action potential is the rapid change of electric potentials on both sides of the cell membrane. In a matter of milliseconds the outer space increases from -90 mV to +20 to +30 mV [4]. Any change of potential that exceeds the threshold level, an action potential occurs.

The mechanism by how an action potential occurs begins by the opening of the sodium channels that exist on the cell membrane. This allows the entry of sodium ions and rapidly increase the electric potential gradient inside the cell, this process is called as depolarization. Since the system strives towards equilibrium an outflow of potassium ions rushes outwards. After an action potential is initiated the potential stabilizes and then falls down below the threshold level, this process is called Repolarization. The channels permeability, namely the sodium potassium pump, is affected by the electric field gradient among other factors. Depolarization can be initiated again after a period of two thirds of the repolarization period, this phase is called the relative refractory period. Since cells are of many types, we are only concerned with myocardial cells that exist in the heart muscle tissue. The refractory period of myocardial cells lasts approximately 100-300 ms and the electric potential is approximately +15 mV [5].

The action potential propagation through the heart starts as an impulse at an origin location and spreads over the entire heart. Heart cells are connected by intercalated discs that have low electrical resistance, thus helps in signal propagation.

2.2.1. ECG signals. Electrocardiogram is a device used to get the surface action potentials of the heart, by employing electrodes at different locations of the body and an amplifier circuit. ECG is a non-invasive way of diagnosing heart related diseases easily by knowing the time, amplitude and direction of different heart signals.

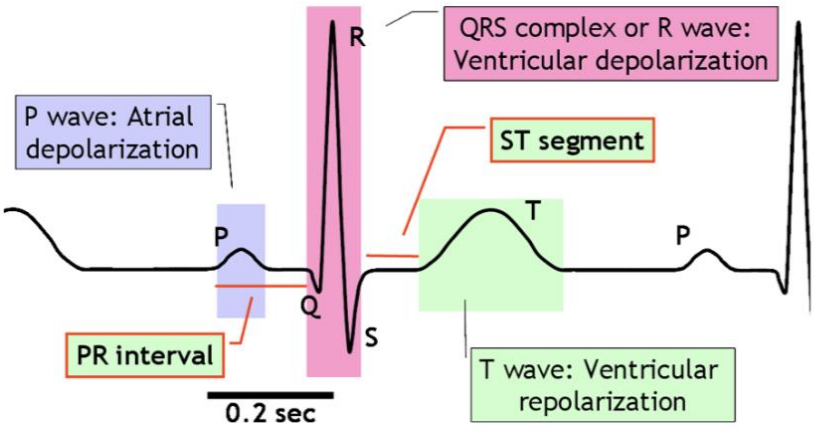

Figure 7: ECG Signal [5]

As shown in Figure 7, ECG signals are typically 2mV in amplitude pk-pk and exist in the range of 0.5-150 Hz [6]. The P signal represents the atrial depolarization, which is followed by a functional delay, where the ventricles gets filled with blood to be ready to pump. Then a huge deflection called the QRS complex happens which indicated the ventricle depolarization. The QRS is followed by a time gap called the S-T interval, which give times for the ventricular repolarization, which simply means returning to its original state. The T wave is a low frequency wave which indicate the ventricular repolarization event taking place.

2.3. Pacemaker Architectures

Pacemakers have seen tremendous changes over the years, we will be discussing their evolution since their inception in this section. However, before we move on we need to understand how pacemakers are made and how they operate.

Pacemakers are electronic devices made to treat any malfunction in the electric system of the heart. The first one was introduced in 1932 and ever since it has continued and will continue to advance[6]. The device is a metal enclosure containing electronics that are connected to probes with electrodes at the end of it in order to sense and stimulate the heart. The first pacemakers used to excite the ventricles asynchronously and did not have the ability of electrogram sensing. The next generation of pacemakers that were called "demand mode pacemakers" included a sense amplifier for measuring cardiac activity to avoid the competition between paced and intrinsic rhythms. In 1963, pacemakers were able to excite ventricles according to atrial excitation [6].

There are several types of pacemakers and the doctor usually chooses the type based on the patient's disease[1], [7]–[12] Sick sinus syndrome and AV block are examples of different diseases that require different pacemakers. A pacemaker usually consists of an electric pulse generator, power source and electrode/s. The output width and duration dictate the amount of output power used to stimulate the heart.

2.3.1. Asynchronous pacemakers. Pacemakers can work in either a unipolar or bipolar lead systems. A unipolar system contains only one probe with an electrode attached at the end. A bipolar system contains 2 separate probes with electrodes attached at the end of it with no more than 15mm distance between them [6].

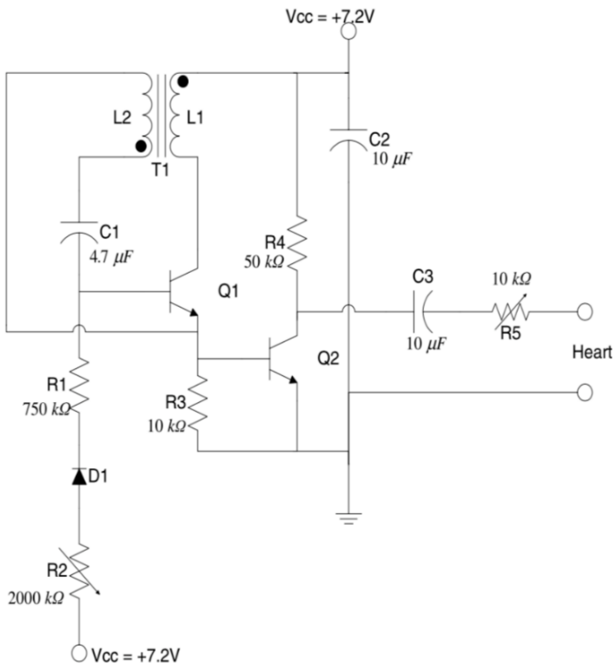

Figure 8: First implanted pacemaker [13]

The first implantable pacemaker was used in 1959, Engineer Wilson Greatbatch and the cardiologist W. M. Chardack developed the first implantable pacemaker [6]. It was used to treat atrioventricular block, the circuits delivered 1ms wide with a pulse amplitude of 10 mA at a frequency of 60 bpm. The current drain from this circuit was calculated to be about 12 μA. It was a single chamber ventricular pacemaker and can be seen in Figure 8 and the functional blocks of a pacemaker could be seen in Figure 9.

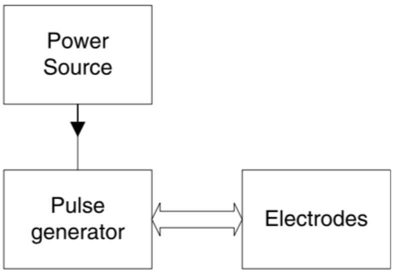

Figure 9: Basic pacemaker block diagram [6]

2.3.2. Demand Pacemakers. Demand pacemaker is another type of pacemakers that follow a very different working principle, as can be seen in Figure 10. The previous pacemaker discussed provide a constant timed pulse to the heart electrical node without taking into consideration the intrinsic nature of the heart electrical system. The demand pacemaker implements a sensing amplifier that takes into consideration the inherent heart behaviour. To sum up, this pacemaker provides a stimulate only in the absence of a natural heartbeat. Demand pacemakers is the basic principle for all modern pacemakers.

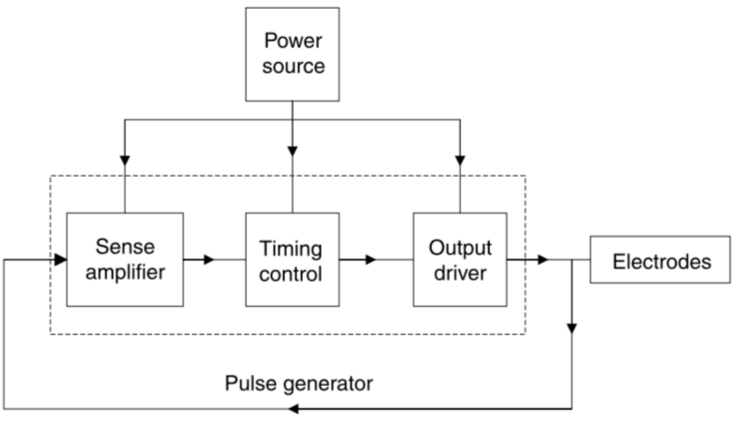

Figure 10: Demand pacemaker block diagram [6]

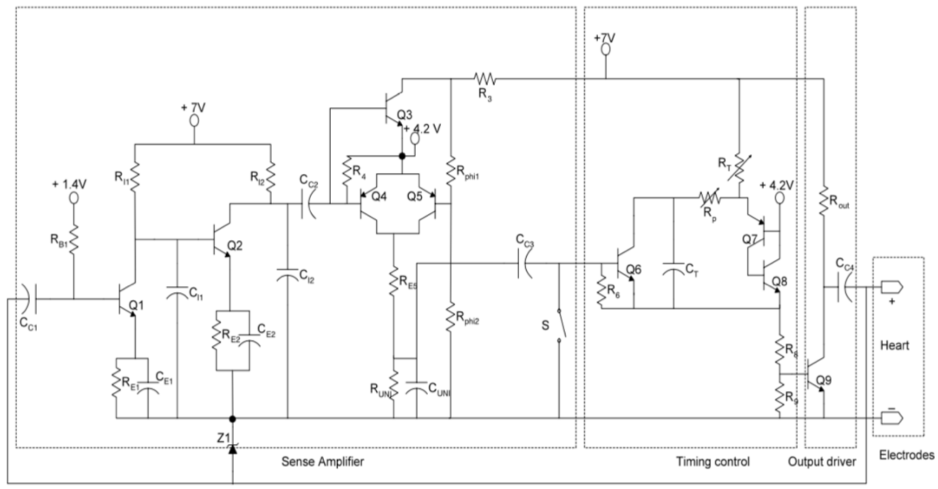

Figure 11: Demand pacemaker circuitry [14]

The sensing amplifier is a major part of the circuit as it blocks pulse generation whenever the heart beats on its own. The amplifier takes in the heartbeat signal, amplifies it and then normalizes it and it filters out unwanted signals such as the P and T waves and the 60 Hz stray signals. The entire circuitry design is shown in Figure 11.

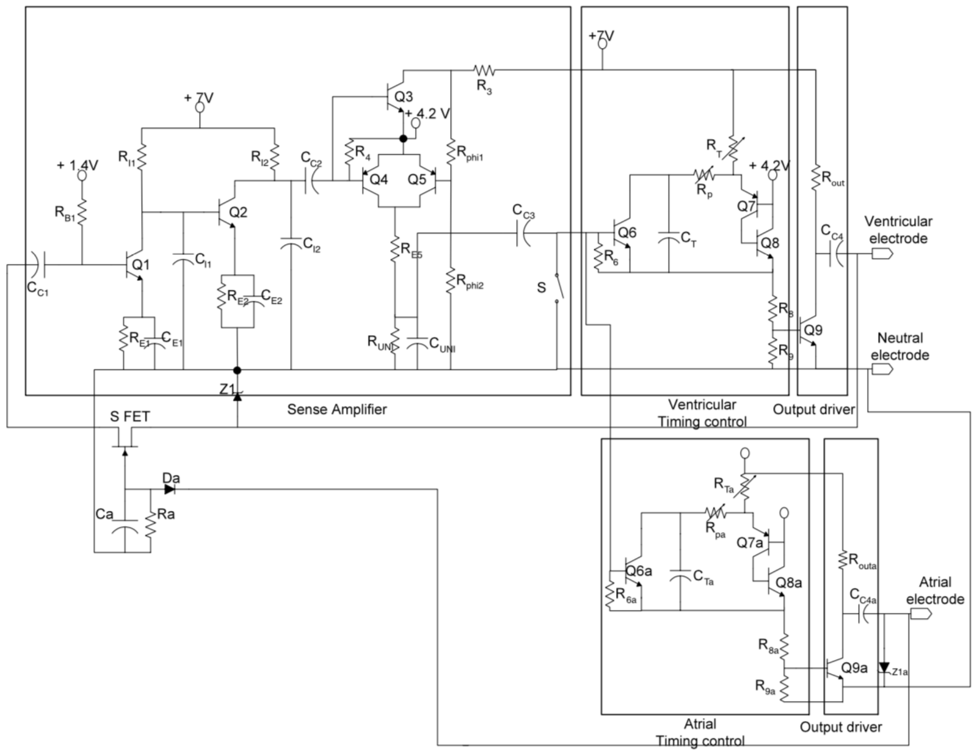

Figure 12: Dual Chamber demand pacemaker circuit [15]

The dual chamber circuit senses the atrial excitation and then stimulates the ventricles. A sensing amplifier was also implemented to detect ventricular intrinsic behavior. The circuitry is also shown in Figure 12.

Rate responsive pacemakers, in Figure 13, were the generation after demand pacemakers. They would regulate the stimulation based on sensor input in the device, such as body motion, pH, respiration, blood pressure - etc.

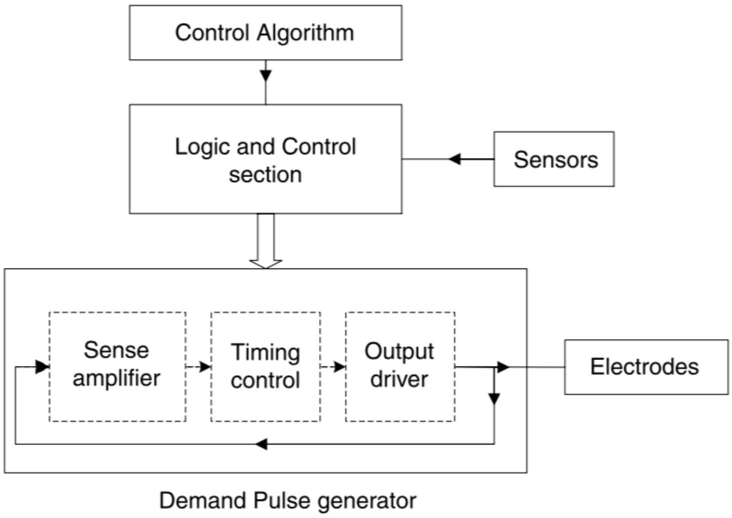

Figure 13: Rate responsive pacemaker diagram [6]

2.3.3. Modern Pacemakers. Modern pacemakers now consist of a telemetry system, analog sense amplifier, analog output circuitry and a microprocessor to act as a controller, shown in Figure 14. This allows for a whole range of features such as real time sensing that can be used to diagnose or monitor current events happening in the heart.

Modern pacemakers have been used frequently in the past decade due to their versatile nature in terms of the capabilities that it offers to patients and doctors. On the patient side, it captures a lot of data from their bodies, with regards to their heart rhythm and other bodily functions such as the respiratory cycles and level of activity of the patient. On the doctor side, it provides critical data that allows them to better provide an improved quality of life to their patients taking into account the vast amount of data that is supplied.

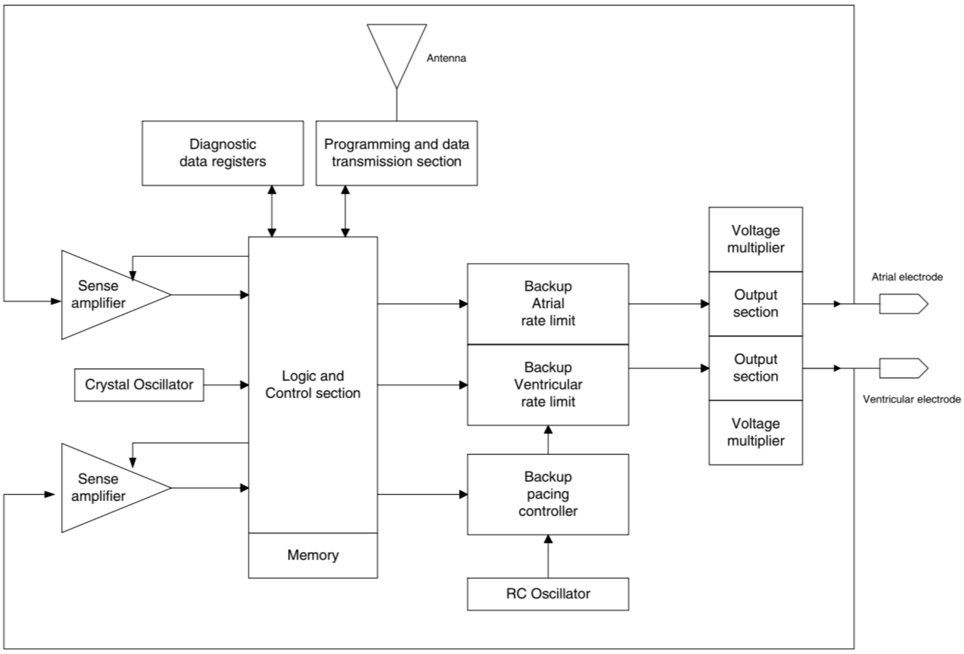

Figure 14: Modern features in pacemakers block diagram [16]

2.4. Pacemaker Modes

Pacemaker codes are divided into 5 positions according to the NBG code. Each position has a special meaning and is referred to by letters. We have in Position 1 indicating which chamber is being paced, and you would find the letters A (Atrium), V (Ventricle), D (Dual, A + V) or O (Off). Position 2 indicates the chamber sensed and it could contain any of the letters found in position 1. Position 3 indicated the response of the pacemaker to its sensing capabilities, and could contain I (Inhibit), T (Trigger), D (Dual, I or T) or O (Off). In position 3, it is important to bear in mind that you should never see the letter T, as it is almost never used. Position 4 indicates if the pacemaker is rate responsive or not, it could contain R (Rate response ON) or Blank (Rate response OFF). Position 5 indicates multi sight pacing, but we will not discuss this as it is not related to our research. Now we have several modes and we will group them in 2 categories, single chamber modes and dual chamber modes. All the modes are shown in Table 1. These mode codes exist in order to properly differentiate between the working modes of different pacemakers and aids in helping medical professionals in choosing the proper type.

Table 1: Pacemaker modes of operation

Single Chamber Modes	Dual Chamber Modes	
VOO	Tracking Modes	Non-Tracking Modes
VVI		
AOO	DDD	DDI
AAI	VDD	DOO

The easiest way to understand how each mode operates is to simply relate each letter to the positions previously mentioned, this is illustrated in Figure 15. VOO, is Ventricular pacing with sensing and response to sensing off. This simply means we have a ventricular pacing device with fixed rhythm that ignores the heart intrinsic electrophysiology. VVI is Ventricular pacing with ventricular sensing and the response of the pacemaker is to inhibit. This means that the device pays attention to the heart intrinsic activity and withholds its pacing when the heart does its job correctly. This prevents abnormalities such as beat fusion and unnecessary power loss. AOO is Atrial pacing with sensing and response to sensing off. It follows the same logic as VOO. AAI is Atrial pacing with atrial sensing and response to sensing set to inhibition.

In dual chambers we can sub-divide it to 2 subcategories, Tracking and non-tracking modes. In tracking mode, we have DDD and VDD. DDD is pacing in both the atrium and ventricles with sensing in both and the response to sensing is inhibition and triggering.

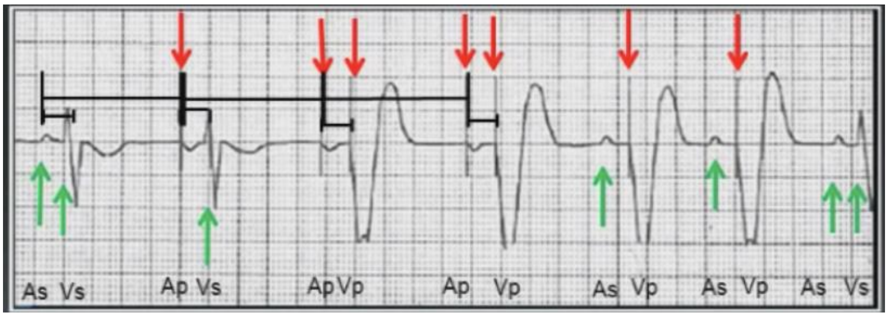

Figure 15: DDD working mechanism [6]

VDD is pacing in the ventricle with the sensing set to atrium and ventricle and response to sensing is inhibition and trigger. The downfall of this design is that at any point if the atrial rate drops below the lower limit, Atrial-Ventricular synchrony will be lost. In non-tracking mode we have DDI and DOO. DDI is atrial and ventricular pacing with sensing in both and response to sensing set to inhibition. DOO is pacing of the atrium and ventricle with sensing and response to sensing off also known as asynchronous pacing.

Rate response is usually used in chronotropic incompetence patients. Chronotropic incompetence is defined by the inability of the heart to adjust its own rhythm to the body's demands, whether it is metabolic or exercise demands. Sensors are utilized to this end and usually two types of sensors are utilized, accelerometer to detect motion and minute ventilation which detects changes in respiratory rate. The data taken is usually frequently plugged into an algorithm which computes the appropriate rate to pace the heart, the algorithm however needs to be optimized as we are not all physically the same.

Choosing a pacing mode is an important design step. It is usually based on cardiac output and AV synchrony. Cardiac output is defined by the stroke volume x heart rate. While AV synchrony is simply the time delay between the atrium and ventricle pacing. AV synchrony must be always maintained and for that the patient must have a viable atrium.

2.4.1. Capture management. Capture management is measuring the pacing threshold and then adjust the pacing output accordingly. Normal impedance for a lead is 300 ohms to 1500 ohms. Threshold is the minimum amount of electrical energy needed to consistently capture the heart outside of the heart's refractory period, it is also known as pacing threshold or stimulation threshold. Threshold contains 2 components, amplitude and pulse width. Safety margin is the margin required to make sure we always capture the heart; it is usually 2 times the voltage or 3 times the pulse width. Sensing is what the pacemaker sees in terms of ECG, or its intrinsic depolarization. Safety margin also exists in the sensing aspect, the standard is 2 times. So in the atrium we can sense half the measured p-wave and in the ventricle we can measure half the measured R-wave.

2.5. Energy Harvesting Systems For Cardiac Applications

Energy harvesting is defined as a method or technique from which we can harvest energy and convert it from one form to another. The human body has all sorts of energy inside it, for example chemical, mechanical and electrical energies. The existence of those energies encourages us to use an energy harvesting mechanism in order to power up our device without the need of batteries. A lot of methods were devised in order to harvest energy for implantable devices using the natural occurring energy inside the human body or providing an external power source. An example would be the use of blood flow, pressure gradients, electromagnetic energy-based harvesters and vibrational energy harvesters [13]–[22]

2.5.1. Energy harvester designs. Different types of piezoelectric designs are discussed in this section. As mentioned previously in section 2.5, energy harvesting modalities have been brought forward since the realization that conventional batteries need to be replaced for low power applications such as wireless sensors and implantable electronics, the latter being focused on heavily in healthcare applications. In this section we will be talking about the different structures in which piezoelectric harvested are designed in cantilever, cymbal, stack and shell type.

2.5.2. Physics of energy harvester. The theoretical science behind piezoelectric energy harvester will be discussed in this section in detail [23]–[30]. These harvesters produce an electrical charge "Q" when it is subjected to a mechanical stress. The charge is then accumulated at the two metal electrodes on the piezoelectric material, therefore creating a voltage across the electrodes. This can be expressed by the equation:

$$v = \frac{Q}{\varepsilon_r \varepsilon_0 \frac{A}{t}} \tag{1}$$

ε0, εr represent the permittivity in vacuum and the relative permittivity between the electrodes across the material, respectively. A and t represent the active area of the material and the thickness of the piezoelectric material, respectively.

We also have the constitutive piezoelectric equation that relates stress-charge.

$$D = dT + \varepsilon^T \bar{E} \tag{2}$$

where D is the displacement vector, d is the electromechanical coupling factor, T is the mechanical stress vector, εT is the dielectric permittivity tensor and E is the electric field vector.

It is important to note that depending on the direction of the mechanical vibration the electric field response will be different due to the anisotropic nature of the piezoelectric material. The authors in [23] recommend that when using any physical property variable of the piezoelectric like for example “d” we should add the subscript 1,2 or 3 depending on the axial orientation.

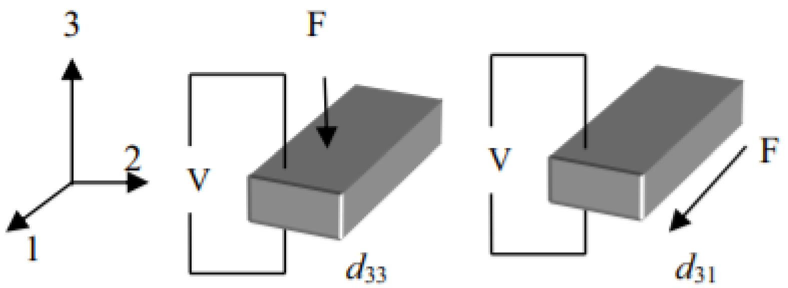

Figure 16: How to label orientation of forces [23]

If we have dij, the first subscript i represents the direction of polarization generated in the material due to the vibrational energy or the direction of the applied electric field. The second subscript “j” denotes the direction of the applied stress of induced strain.

We will now discuss one of the most important equations required in any piezoelectric design, the resonance frequency. It is optimal to have the resonance frequency of the system to be the same as the device in order to yield the highest possible energy harvesting.

$$f_r = \frac{v_n^2}{2\pi} \frac{1}{L^2} \sqrt{\frac{D}{m}} \tag{3}$$

Where f_r is the resonance frequency, v_n is the resonance frequency in radians, *L* is the length, *D* is the electric displacement and *m* is the mass per unit area.

$$m = \rho_p t_p + \rho_s t_s \tag{4}$$

ρ_p is the density of the proof mass, ρ_s is the density of the substrate, t_p is the thickness of the proof mass and t_s is the thickness of the substrate.

However in [24], it is defined in a different manner.

$$W_n = \sqrt{\frac{K}{m}} \tag{5}$$

where W_n is the resonance frequency, *K* is the spring constant, *m* is the mass per unit area.

The computed power for the electrical energy acquired is defined as:

$$P_e(max) = \frac{mA^2Q}{4W_n} \tag{6}$$

where $P_e(max)$ is the electrical energy power, *A* is the amplitude of the signal and *Q* is the quality factor.

2.5.3. Cantilever types. A cantilever type energy harvester is a very simple structure that can produce a very large deformation under vibrations. There exists different types of configurations for this system such as unimorph, bimorph series and bimorph parallel

2.5.4. Harvesting modes. The harvesting modes can be in the 31 or the 33 mode. However, 31 mode is the most useful in harvesting applications because of the mass utilization. In unimorph cantilevers, the maximum power density is achieved when the vibration frequency matches the resonant frequency of the device itself. Unimorphs are designed to be in the range of 60-200 Hz.

2.5.5. Unimorph cantilever equations. In this section we will be providing the theoretical equations needed to calculate the design parameters for a unimorph cantilever piezoelectric device [21].

$$f = \frac{\omega_n^2}{2\pi} \sqrt{\frac{0.236 D_p w}{\left(l - \frac{l_m}{2}\right)^3 (m_\epsilon + m_p)}} \tag{7}$$

where *f* is the resonance frequency in hertz, ω_n is the frequency in radians, D_p electric displacement, *w* is the width of the beam, l is the length of the beam, l_m is the length of the proof mass, m_ϵ is the effective mass of the cantilever and m_p is the proof mass.

$$m = \rho_p t_p + \rho_s t_s \tag{8}$$

where m is the mass per unit area, ρ_p is the density of the proof mass, ρ_s is the density of the substrate, t_p is the thickness of the proof mass and t_s is the thickness of the substrate.

$$D_p = \frac{(E_p^2 t_p^4 + E_s^2 t_s^4 + 2E_p E_s t_p t_s \left(2t_p^2 + 2t_s^2 + 3t_p t_s\right))}{12(E_p t_p + E_s t_s)} \tag{9}$$

where E_p is the young's modulus of piezoelectric material, E_s is the young's modulus of substrate.

$$Q = \frac{-3 d_{31} s_s s_p t_s \left(t_s + t_p\right) l^2 F}{B} \tag{10}$$

Q is the quality factor, d_{31} is the dielectric coefficient in the direction of 31, s_s is the elastic compliance of substrate, s_p is the elastic compliance of the proof mass, *F* is the force applied and *B* is the bandwidth.

$$s_s = \frac{1}{E_s} \tag{11}$$

$$s_p = \frac{1}{E_p} \tag{12}$$

$$B = s_s^2 t_p^4 + 4 s_s s_p t_p^3 t_s + 6 s_s s_p t_s^2 t_p^2 + 4 s_s s_p t_p t_s^3 + s_p^2 t_s^4 \tag{13}$$

$$s_h = s_s t_p + s_p t_s \tag{14}$$

where s_h is the total elastic compliance.

$$V = \frac{-3d_{31}s_s s_p t_s t_p(t_s + t_p)lF}{\varepsilon_{33}^T wB(1 + \left(\frac{3s_s s_p^2 t_s^2 t_p\left(t_s + t_p\right)^2}{s_h B} - 1\right)K_{31}^2)} \tag{15}$$

V is the voltage, ε_{33}^T is the dielectric permittivity in the direction of 33 and K_{31} is the spring constant in the direction of 31.

Here we will introduce the stain-charge form equations for the piezoelectric:

$$S = s^E T + d_t E \tag{16}$$

where *S* is the strain, s^E is the elastic compliance, *T* is the mechanical stress, d_t is the piezoelectric coefficient and E is the electric field.

$$D = d_t T + \epsilon^T E \tag{17}$$

where D electric displacement.

2.5.6. Different energy harvesting techniques. As shown in Table 2, researchers have discovered a wide range of techniques in order to harvest electricity from naturally occurring phenomena. Table 2 shows a comparison between piezoelectric, electrostatic and electromagnetic approaches from different criterions.

Table 2: Difference energy harvesting mechanisms comparison table [20]

Features	ENERGY HARVESTING TECHNIQUES		
	Piezoelectric	***Electrostatic***	***Electromagnetic***
Separate Voltage source	No separate voltage source is required	Can be easily implemented as a MEMS	No separate voltage source
Energy Density	Highest energy density	A decrease in the separation between capacitor plates increases the energy density	Energy density is low
Output	Outputs obtained are approximately (2-10) V	Outputs obtained are approximately (2-10) V	Outputs only up to 0.1 V
MEMS Implementation	Difficult to integrate with MEMS	Easy to be implement in MEMS	Difficult to integrate with MEMS
Frequency Dependence	Highly frequency dependent	No dependence	No dependence
Issues	Possible bio compatibility issue	Requires initial polarizing charge	Possible issues with other medical equipments such as an MRI
Other features	Materials can be expensive	Electrets can be used to provide the initial charge needed)	Bulky in size

2.6. Energy Storage

This design is concerned with the use of energy harvesters in pacemaker implantable devices, the storage of energy is crucial as it facilitates an uninterrupted power supply to the pacemaker's electronics. A multilayered substrate can be used where a switched capacitor power converter could be used beneath the energy harvesting unit, accommodating the unused space. This type of capacitor system is used in low-power, millimeter applications for its fine power control and easy chip integration [26]. In the system shown in Figure 17, the entire system dimension is around 2 mm^2 with a wide operation range and low quiescent power. The overall conversion efficiency is also acceptable at this low power level.

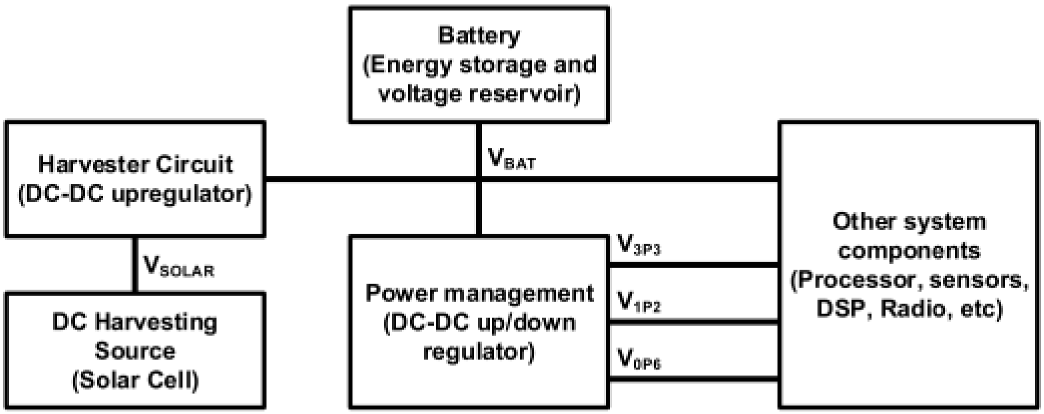

Figure 17: Energy storage system block diagram [30]

Chapter 3. Methodology & Experimental Setup

In this chapter, the problem of developing an energy harvester system using COMSOL Multiphysics package is discussed. An implementation of the system prototype is described. The simulation was done on COMSOL Multiphysics package.

3.1. Problem Formulation

In order to design an energy harvester, we now have the proper understanding of the physiological parameters and the technological requirements to realize such a system. The aim of this research is to design a low frequency bimorph cantilever energy harvester for pacemaker application. The system needs to be designed with the application of pacemakers in mind.

3.2. Proposed Simulation Software

The simulation of the energy harvester will be done on COMSOL Multiphysics package. COMSOL offers a wide range of tool to accurately simulate a device, considering the numerous physical phenomena occurring in the structure. COMSOL also has a very friendly graphical user interface, which makes the data visualizing a unique experience. It also had the ability and tools to optimize and edit individual parameters or even multiple ones at once.

3.3. Physical Design Parameters For Resonance

The resonant frequency of the cantilever bimorph energy harvester needs be set at a low frequency in order to accumulate as much power from the heart as possible to this end we will be introducing the equations necessary for that design and analysis of the system.

3.4. Implantation Technique

Leadless pacemakers are not a new concept, in fact it is a well investigated subject, which has developed into available products that patients can readily use [18]. Moreover, systems for patients with severe arrythmia disorder were based on a capsulated pacemaker system. A transcatheter procedure could effectively implant the pacemaker system inside the region of interest in the heart without the need of the direct invasiveness of surgery.

3.5. Fabrication Process

The fabrication method of the energy harvester, that will be soon discussed in details, can be realized using commercially available products. A common PZT 5A wafer of 0.5 inches diameters and 0.0075 inches thickness costs about 90 united states dollars. A bimorph stack from piezosystem can be used to build the PZT-5A bimorph layer sandwiching the aluminium substrate and base. The proof mass is going to be a block of aluminium at the end of the cantilever system. The maximum length of the cantilever system is approximately around 21mm encapsulated in a biocompatible material to be safely working in the body environment without risking any attack from the patient immune system. The final prototype can be simulated with a shaker vibrating at the resulting resonance frequency with the same vibrational intensity of the heart, where the voltage and output power can then be measured.

3.6. Bimorph Cantilever Equations

Stoney's equation can be used to relate the applied force σ to the cantilever end deflection δ:

$$\delta = \frac{3\sigma(1-v)}{E\left(\frac{L}{t}\right)^2} \tag{18}$$

In equation (18), *v* is the Poisson's ratio, *E* is young's modulus, *L* is the cantilever's length, *t* is the cantilever's thickness. The cantilever spring constant *k* can also be expressed as:

$$k = \frac{3EI}{L^3} \tag{19}$$

where in equation (19), *I* is the moment of inertia of the cantilever and is also expressed by:

$$I = \frac{bh^3}{12} \tag{20}$$

In the above equation (20), b is the width of the cantilever and h is the thickness. The resonant frequency is then calculated based on the general equation:

$$f = \frac{0.32\sqrt{k}}{\sqrt{m}} \tag{21}$$

In the above equation (21), f is the frequency at resonance. The power that is being generated can be defined as:

$$|P| = \frac{RC^2\left(\frac{Ydg}{k\varepsilon}\right)^2}{(2\xi\omega^2 RC)^2 + (\omega^2 RC(1-k) + 2\xi\omega)^2}\frac{3b}{2l^2}|A|^2 \tag{22}$$

where R is the resistive load, C is the capacitance of the piezoelectric, Y is the displacement magnitude of input vibrations, ξ is the damping ratio, k is the piezoelectric coefficient, ω is the frequency, l is the length of the beam, ε is the electric permittivity, d is the piezoelectric strain coefficient, g is the force of gravity and A is the acceleration magnitude of input vibrations [31].

3.7. Cantilever Structure

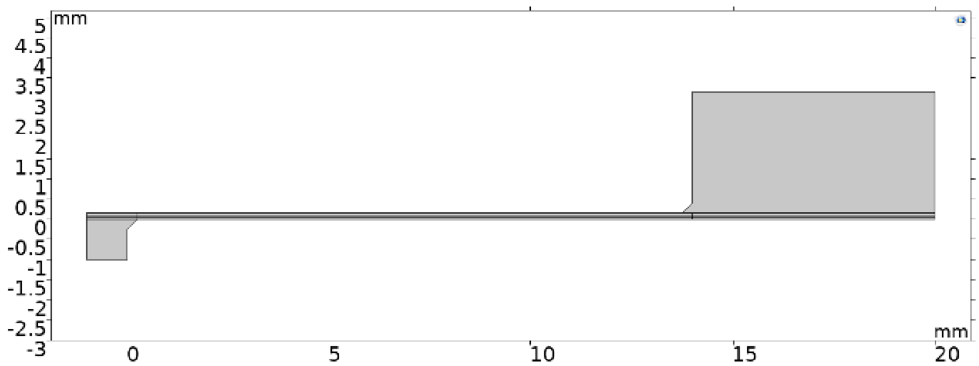

Figure 18: Cantilever Energy Harvester Design

Figure 18 shows the simulated design with dimensions of 21mm width, 0.16mm thickness and 10mm out of plate dimension. There are three layers with thicknesses of 0.06mm for the PZT layers and 0.04mm for the supporting layer [32]. A proof mass was added to the cantilever to lower the resonance frequency with dimensions of 6mm width and 3mm height. The structure is then boundary fixed on 1mm by 1mm support at the far-left corner.

The materials used in this simulation are PZT-5A and structural aluminium with properties shown Table 3. The properties shown pertains to the density, Poisson' Ratio and Young's Modulus.

Table 3: Materials' Properties

	Materials	
Material Names	PZT-5A	Structural Aluminum
Density (kg/m^3)	7750	2700
Poisson' Ratio	0.31	0.35
Young's Modulus (Pa)	63e9	70e9

3.8. Simulation Conditions

Boundary conditions, circuit terminals and ground were considered in the design, as well as the damping and mechanical properties of the materials used. The simulation type is frequency response of the entire structure sweeped at a frequency range of 30Hz to 70Hz. The mesh settings were set to triangular mesh with maximum element size of 0.02mm and minimum element size of 0.002mm. The acceleration was set to 1g and the Resistive load $12k\Omega$.

3.9. Testing Parameters

The simulation results contain 5 sections, the von Mises stress (N/m^2), Electric potential, frequency response, Load dependence and acceleration dependence. The reason for providing all these testing parameters is to properly have more information about the system when put into real life application. The von mises stress analyses will measure longevity. The electric potential analysis will visualize the voltage produced from the system. The frequency response, acceleration dependence and load dependence will exhibit how the system responds to vibrations of the heart while also analysing the resonance of the system.

Chapter 4. Results and Analysis

In this chapter, we present the experimental results for the COMSOL simulation and testing.

4.1. Simulation Model Results

A simulation that yielded a resonant frequency of 48Hz with a maximum electrical power output of 4.77 mW for an input mechanical power of 4.88 mW was obtained.

4.2. Testing Parameters Results

This section will discuss in details the results obtained as well as the analysis of it in the five main categories of testing which are the von mises stress, electric potential, frequency response, load dependence and acceleration dependence.

4.2.1. Von mises stress. The von mises stress is a measure of the structure yielding stress, this is very important when durability is in question. Figure 20 shows the structure's deflection at resonance frequency, which is equivalent to the maximum deflection done by the cantilever. Based on the results shown above, the maximum stress that will be exerted by the structure is well below the material fracture stress by a significant amount, this directly correlates to the durability of the structure. The x and y axis are labelled in millimetres.

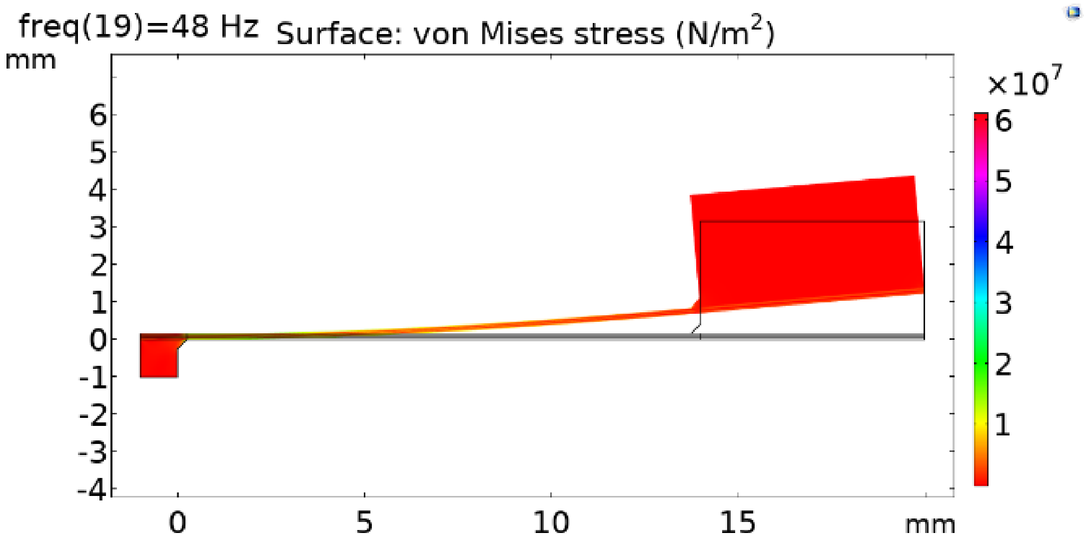

Figure 19: Von mises stress

4.2.2. Electric potential. While looking at a cross-sectional area of the beam as shown in Figure 20, we can see the induced electric potential at maximum deflection. The voltage yielded is 10.7 Volts. The y-axis and x-axis are labelled in millimetre.

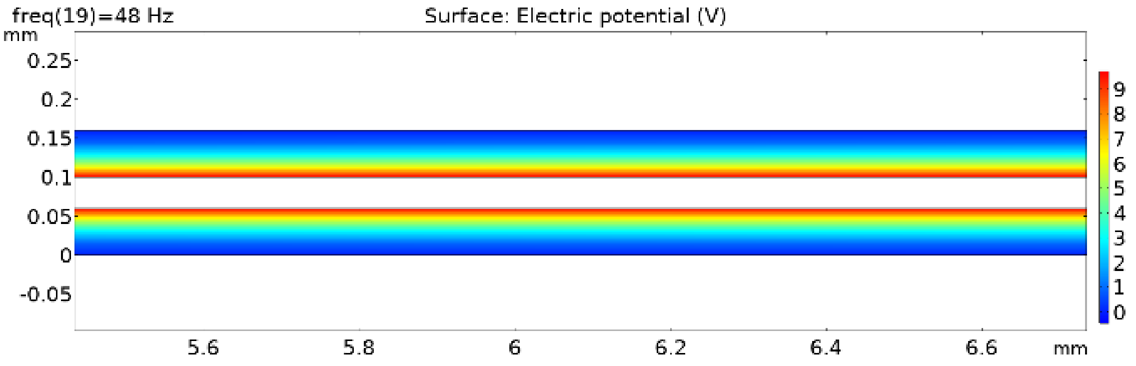

Figure 20: Electric potential induced at resonant frequency

4.2.3. Frequency response. In this step, as shown in Figure 21, the frequency was sweeped from 30Hz to 70Hz at a step of 1Hz per iteration, the electrical power output and corresponding mechanical power input were then calculated. To read the graph properly, the x-axis is the frequency in hertz. The y-axis labels depend on the legends shown in the upper right corner of the graph, if we concerned with voltage for example, the blue line shows the trend of the voltage against the frequency, therefore the y-axis becomes the voltage in volts.

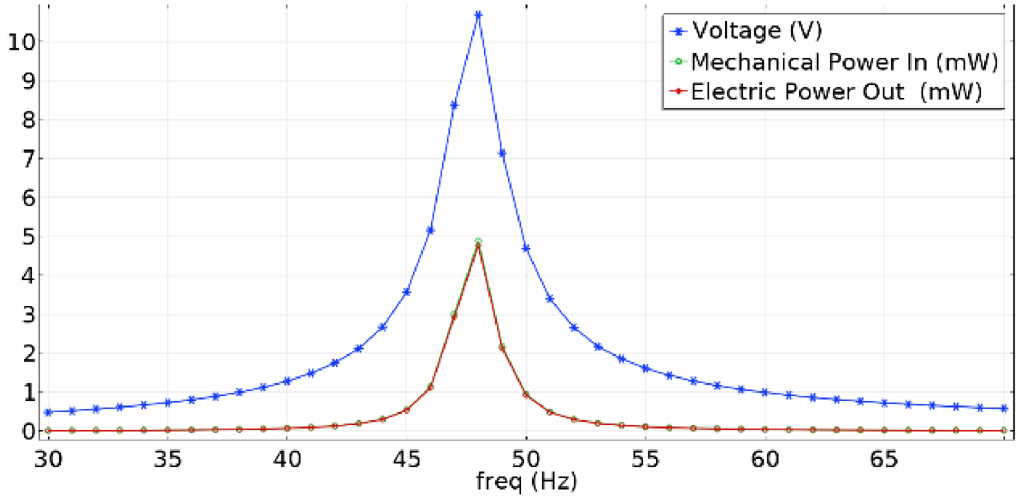

Figure 21: Frequency response

4.2.4. Load dependence. As observed in Figure 22, the voltage of the system increases as the resistive load increases from 100 Ω to 100 $k\Omega$. But at 5600 Ω the power output drops significantly; this is because the load needs to be matched to the overall system. The x-axis is labelled as the resistance in ohms, the y-axis label depends on which curve is being observed depending on the legend shown in the upper left corner.

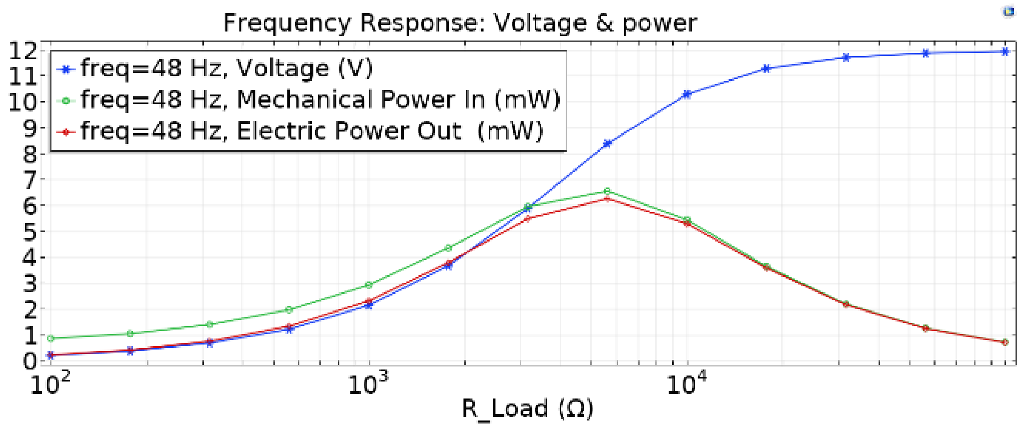

Figure 22: Load dependence

4.2.5. Acceleration dependence. The acceleration set for the simulation was 1g as the human body is objected to that g-force. However, if the g-force was varied the voltage and power in and out were calculate as shown in Figure 23. Figure 23 has the acceleration on the x-axis and the y-axis represents the voltage, mechanical power in and electrical power out in milli watts. This shows an expected increase in the voltage and power performance of the device as it would increase the stress forces on the beam. The x-axis is labelled as acceleration in g force, the y-axis label depends on which curve is observed as shown in the legend in the upper left corner.

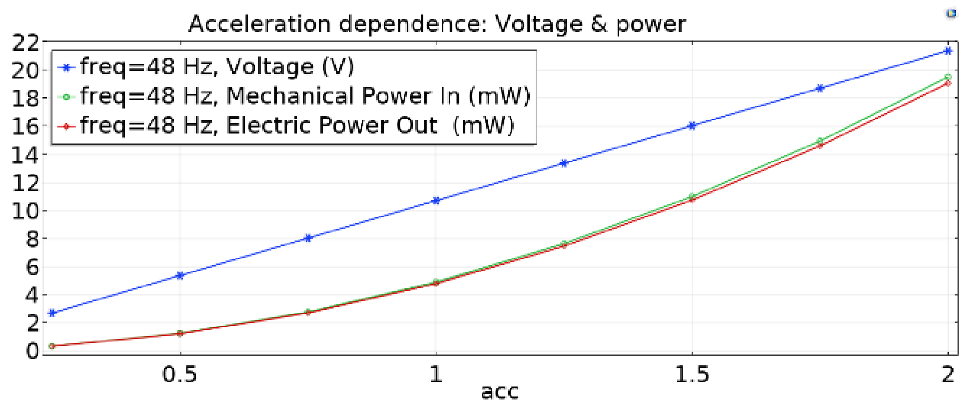

Figure 23: Acceleration dependence

Chapter 5. Conclusion and Future Work

The motivation behind this thesis topic is that cardiovascular diseases is one of the leading causes of death around the world. Arrythmias is one of the most common diseases that are heavily discussed under the umbrella of cardiovascular diseases. While existing pacemakers provide a long term and reliable solution to many heart illnesses, the need to improve on them is needed to provide a better quality of life to the patients.

In this thesis, a bimorph cantilever energy harvester design that used PZT-5A and structural aluminium as construction materials was presented. The performance of the device was evaluated in terms of frequency response, voltage, mechanical power input and electrical power output. The durability of the model was investigated using the von mises stress calculated which was obtained using COMSOL.

The simulated design has yielded 10.7 Volts, 4.77 mW electrical power for a 21mm wide cantilever beam, which is suitable for implantation in the heart using a transcatheter intervention based on the information provided in the literature [22]. The dimensions are also suitable for the application of leadless or self-powered pacemakers which would help power the device using the patient's heartbeat as a power source.

Power storage techniques were studied in order to view how this design would integrate with a future leadless pacemaker. This was proposed using a low power switched capacitor converter system in the literature [8]. This system was done for low power IOT applications, in order to aid the power storage for the associated energy harvester in the device. The power storage technique will be further investigated and designed in the future work.

Deployment of the device was also discussed to further emphasize the usefulness of small size, energy harvesting pacemakers. A transcatheter intervention is the most favourable form of implantation due to the fact that it is not as invasive as other forms of surgery, namely open-heart surgery, therefore it does not carry a lot of risks and complications.

In conclusion, this work reflects the importance of having a power source alternative for pacemakers to increase their longevity. The importance of it relies in the fact that patients with heart diseases need a better quality of life and that translates to being able to implant the device with minimum invasiveness and with a guarantee that a revision surgery is not needed. In most cases, patients need revision surgery in order to replace the battery of the device or the entire device as a bulk. The inclusion of an energy harvester eliminates that concern.

The positive outcomes of the energy harvester discussed can be summarized into:

- Device power longevity
- Ease of implantation

Device power longevity with regards to the continuous harvesting of energy from the heart muscle itself, rendering the usage of battery useless. Ease of implantation refers to the use of a non-invasive procedure for implantation of the device.

Furthermore, the future work for this project should include the fabrication of this device by contacting relevant foundries and ordering the required materials. A bench testing procedure will be formulated to evaluate the real performance of the device. Also, the energy storage circuit should be further investigated and designed for our system. Moreover, a complete system integration of the energy harvester and a pacemaker will be designed and packaged in a way that allows a transcatheter implantation.

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Vita

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