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Design Concept of a Bio-Electronic Heart using Artificial Muscle

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Abstract: The objective of this study paper is to introduce a new preliminary design of a bio-electronic heart which will facilitate the use of artificial muscle and an electric pulse generator which will enable the structure to act-alike the biological heart. The following research on this design will aim at creating a solution for less heart transplant donors available, transplant rejection and following immunosuppressant medication given to suppress the contradictory reactions between donors and recipients tissues.

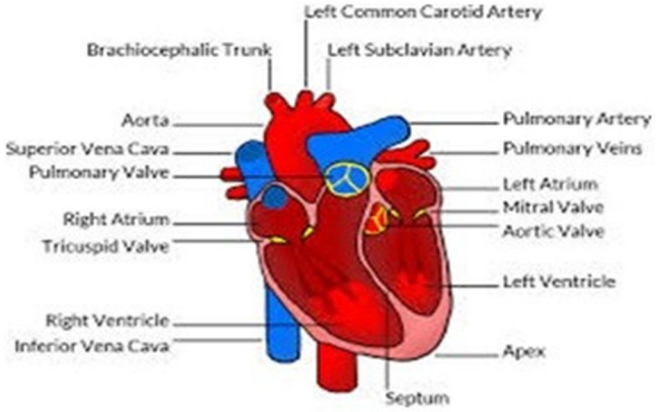

Figure 1.Anatomy

I. INTRODUCTION

The heart is a major organ that keeps us alive and is a symbol of active life. It keeps beating and completing the fundamental cycles of circulatory system-respiratory chemistry, and waste management. But when it gets damaged or stops functioning, it calls for the critical scenario that is today the second most common cause of death. There has been multiple alternative workarounds, in particular, by replacing the damaged parts by engineered vascularized cardiac tissue [1].This paper aims to collaborate both the activities to imitate biological muscle movements of human heart and reproduce them by means of artificial muscles made up of EAPs. Electro active polymers (EAPs) are materials that change their shape and/or dimension in response to an electric stimulus, and thus accomplish movements that are smooth enough to mimic the biological muscles.

II. NORMAL HUMAN HEART

A. Anatomy and Physiology of Human Heart

Two types of circulation: Pulmonary and Systemic Pulmonary circulation involves the movement of blood from the heart to the lungs for carrying out the process of oxygenation and then bringing the blood back to the heart again. In Systemic circulation, oxygen-depleted blood enters the right atrium through the superior and inferior vena cava. The blood is then pumped through the tricuspid valve into the right ventricle; from where it is then pumped into the pulmonary artery through the pulmonary valve. The pulmonary artery splits into the right and left pulmonary arteries, each of which travels to left and right lung. This blood is oxygenized at lungs, as it travels through capillary beds on the alveoli. This is where respiration or the change of gasses occurs, thereby removing carbon dioxide and adding oxygen to the blood. The surface for gas exchange is provided by air sacs called alveoli in lungs. The oxygenated blood is then carried by pulmonary veins, which returns it to the left atrium, thereby completing the pulmonary circuit. Once it reaches the left heart, it flows into the left ventricle through the bicuspid valve from where it is pumped through the aortic valve into the aorta. Systemic circulation begins again thereby delivering oxygenated blood to the body before returning again to the pulmonary circulation.

Additionally the conducting system of the heart consists of cardiac muscle cells and specialized conducting fibres which initiate impulses and conduct them rapidly through the heart. The conducting system provides the heart its automatic rhythmic beat. For the heart to pump efficiently and the systemic and pulmonary circulations to operate synchronously, the events in the cardiac cycle are coordinated. This pathway is made up of 5 elements:

1) The sino atrial (SA) node
2) The atrio ventricular (AV) node
3) The bundle of His
4) The left and right bundle branches
5) The Purkinje fibres

B. Biological and Bio-mechanical models of Human Heart

There has been a lot of work been done in mathematical modeling of normal heart and also that of defected heart[2]. Following are the equations from such work, on the basis that even though blood is non-Newtonian fluid, but can be considered as Newtonian fluid which is governed by the Navier Stokes equations.

$$\frac{\partial Q}{\partial t}+\frac{3Q}{V}\frac{\partial V}{\partial t}-\frac{2LQ^2}{V^2}\frac{\partial V}{\partial z}+\frac{4\pi\mu L}{V}Q+\frac{V}{2\rho L}\frac{\partial P}{\partial z}=0 \quad (1)$$

This master equation (1) is then used to produce the heart chamber mathematical model, which is represented by this second equation (2) from the material [2]

$$\frac{dV}{dt}-\frac{V^2}{6L\rho Q}\frac{\partial P}{\partial z}-\frac{4\pi\mu L}{3}=0 \qquad (2)$$

where the variables hold following values:

- L = 4 cm
- ρ = 1.057 ;(1.043 to 1.057 g/cm^3)
- Q = 3000⁄60; (1 to 5.4 (liter/minute = $1000cm^3/60s$))
- $\partial P/\partial z$ = 70*(10+1325⁄76); (100 to 40 mmHg into g/s^2cm)\
- μ = $0.035cm^2/s$

The closed solution to the above differential equation (2) is as follows:

$$V_h = L\sqrt{\frac{8\pi\mu\rho Q}{\partial z}}\tan\left(\sqrt{\frac{2\pi\mu}{9\rho Q}\frac{\partial P}{\partial z}}\,(t-c_o)\right) \qquad (3)$$

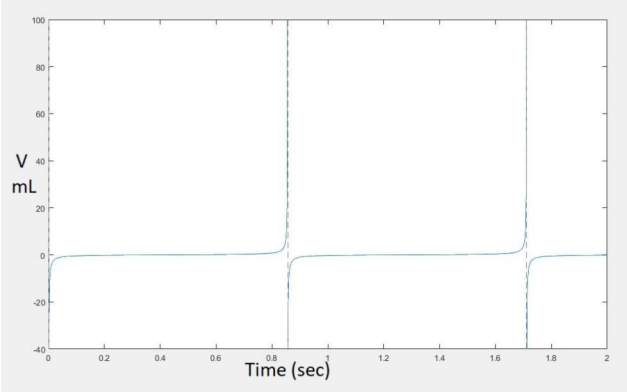

Fig. 2. Plot of volume flow in Heart

The paper shows successful match when used to simulate heart rhythm from real values. These differential equations can be used as the standard to produce analytically similar heart equation for the bio-electronic heart that is proposed in this paper. The paper uses boundary conditions that can be used for bio-electronic heart specifications to build upon.

The pressure generated by EAPs in bio-electronic heart will tend to quantify the natural pressure applied by the chamber walls. This is used to derive an inverse solution by means of a mathematical model which will approximate the designs of the supporting artificial muscles.

Even sophisticated models of human cardiovascular system is stated in this work [3] .

C. Timing Analysis of Systole and Diastole of normal human heart

Cardiac cycle has two parts: Diastole and systole. The lub dub sound as heart beats which we hear are due to blood being pumped through a system of blood vessels. Systole occurs when the heart contracts and blood is pumped out, and diastole occurs after contraction, when the heart relaxes.

For Systole, time for atrium is of 0.7s and 0.5s for ventricles.

For Diastole, time for atrium is of 0.1s and 0.3s for ventricles.

III. ARTIFICIAL MUSCLE

A. History and Anatomy

1) Artificial Muscles are materials that, upon application of electrical stimulus, respond mechanically with significant changes in shape and size. Artificial Muscles are classified on the basis of actuation- EAP (Electro-Active Polymers[4]), Pneumatic, Hydraulic, Thermal, Electrostatic and Electromagnetic. EAPs can be further sub-classified into Electronic EAP and Ionic EAP; Electronic, being operated on higher voltages than Ionic and has slower response time comparatively[7]. In 1990s, two more discoveries of BGA (Bucky Gel Actuator) and IPMC (Ionic Polymer-Metal Composites) were made. Out of these, our concern is with IPMC, nearly uniform curvature on bending and is biocompatible, as it has the maximum compatibility rate with our proposed design[5]. The structured analysis which was carried out, concludes following constraints-

2) Faster response

3) Lower activation voltages[12]

4) Applicable bending displacement.

5) Conductivity depends linearly on the diffusivity of ions. Independent of its dielectric constant

B. Research work in the Area

IPMC consists of a polyelectrolyte membrane, on which a noble metal is plated on both the faces. To balance the balance the electrical[11] charge of the anions, it is neutralized with certain counter ions that are covalently fix to the membrane[9].Ion exchange capability and the water uptake are the two mandatory criterias used for base-polymer selection for IPMC manufacturing; on the basis of which, Nafion[16]. Flemion, Aciplex etc are popularly used[10]. With increase in applied voltage, the bending displacement as well as the generative tip force ted to increase [6] and lower frequencies lead to higher displacement[8].

IV. PROPOSED MODEL OF HEART

A. Associated Stem Cells Research

Cells that have the ability to continuously divide and develop into other kinds of cells are called as stem cells. While the fields like tissue engineering and regenerative medicine derive their roots from stem cells engineering; it is one field that holds promise for treatment of diseases which was never thought before. There are five types of stem cells:

1) *Totipotent:* Can differentiate into embryonic and extra embryonic cells

2) *Pluripotent:* Can differentiate into all types of cells

3) *Multipotent:* Can differentiate into those cells which belong to one family.

4) *Oligopotent:* Can differentiate into few types of cells.

5) *Unipotent:* Possess the property of self-renewal

Usage finds application to mimic biological processes outside the body to replace a disease or damaged tissue inside the body. Generally, groups of cells make and secrete their own support structures, called extra-cellular matrix or the ECM. This ECM has functions like cell adhesion, cell-to-cell communication and cell differentiation.

This matrix called scaffold, does more than just to support the cells. It can be thought of as a network of non-living tissue composed of proteins and carbohydrates to which cells adhere and grow on. Once scaffold is created, cells with or without growth factors can be introduced. Suitable environment is prepared for development of a tissue and in some cases, the cells, scaffolds, and growth factors are all mixed together at once, allowing the tissue to self-assemble. Medical researchers in field of tissue engineering are trying to develop a bio artificial heart, grown from the patient's own cells; an organ like that the body would recognize as its own and thus will not reject.

Process: The preferred method is to remove all cellular components from a heart i.e. to decellularize it. This technique involves perfusing the heart with SDS (sodium dodecyl sulphate), distilled water and Triton X-100. What remains is the ECM composed of structural elements such as collagen, laminin, elastin and fibronectin that helps the living cells grow and stay in just the right place. Because minimal DNA material is left after the decellularization process,the engineered organ is biocompatible with the transplant recipient. The next step is recellularization process. When Hearts are recellularized, they are less immunogenic and also have a decreased rejection rate. The MCPs (monocyte chemoattractant protein) are added to the decellularized heart and with additional exogenous growth factors, are stimulated to differentiate into cardiomyocytes, smooth muscle cells and endothelial cells. Cells which come into picture are endothelial cells, Smooth muscle cells, Fibroblasts & myofibroblasts, genetically modified cells and stem cells. Scaffold here is formed from three different patterns, Synthetic (PET, ePTFE, PGA, PLA, PU), Natural (collagen) and from Decellularized biological matrices. When scientists performed experiments with recellularized rat hearts, the heart models were stimulated by an electrical signal to provide pacing after 8 days of maturation. The heart models showed the activities which was only 2 percent as that of a normal rat heart; which could not support the blood-pumping function.

B. Theoretical Structure and it's Timing Analysis

The design includes making two set of tissue sacs out of scaffold which will act as a two artificial ventricles, of ellipsoid shape; one pumping blood towards lungs for oxygenation and another pumping blood to aorta for distribution throughout the body. The design includes patients myocardial cells injected into mesenchymal stem cells (MSCs) and as stated in the above theoretical section and thus will ensure to eliminate the chances of body rejection. For pumping mechanism, already prepared interconnected nodal structure of artificial muscle will then be attached to this sac tissue; analysis of which is done in the next section. This structure will function on the similar concept of biological heart. It will have the artificial Sino -atrial node, which will be triggered by the internal pulse circuit. It will exhibit the property of contraction when triggered and the impulses through it will travel throughout the nodal structures, thereby making a wave of contraction on the entire ventricle. This contraction will pump the blood to lungs for oxygenation. By the time oxygenated blood returns to heart, the wave of contraction would have reached another ventricle and thus by means of same functionality, blood will be pumped through aorta to process systemic circulation. Now the problem of making a IPMC (Fig. 3) that work in our case becomes an optimization problem :

1) The response time is proportional to thickness of the strip[14]. ($t_r \propto t$)
2) The response speed is inversely proportional to strip thickness. ($s_r \alpha 1/t$)
3) The tip displacement is also inversely proportional to strip thickness ($d_{tip} \alpha 1/t$)
4) But the force is proportional to the square of the strip thickness[10].

$$F = C \times \frac{wt^2}{L} \quad (4)$$

The threshold voltage in case of ionic EAPs (IPMC) is lesser than that required in actuation of the counterpart electronic EAPs. And the deviation as can be seen(Fig. 4).

$$\delta = \frac{FL^3}{3YI} \quad (5)$$

Y is the young's modulus of the strip, I is the moment of Inertia of the strip. Considering rectangular cross-sectional beam

$$I = \frac{wt^3}{12} \quad (6)$$

hence,

$$\delta = \frac{4FL^3}{Ywt^3} \quad (7)$$

Now from the torque formula from [13]

$$\tau = FL = \frac{L'}{k} t^3 E \qquad (8)$$

It can be readily approximated that

$$F(x) \sim \frac{L'}{k} \frac{t^3 E}{x} \qquad (9)$$

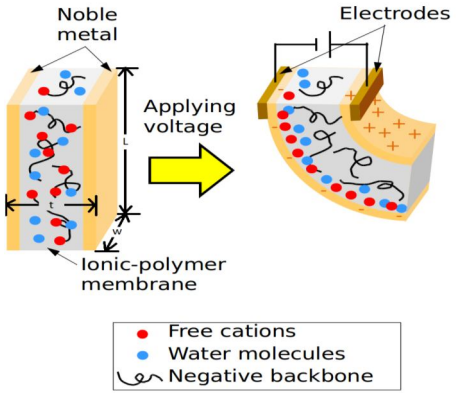

Fig. 3. Schematics of IPMC

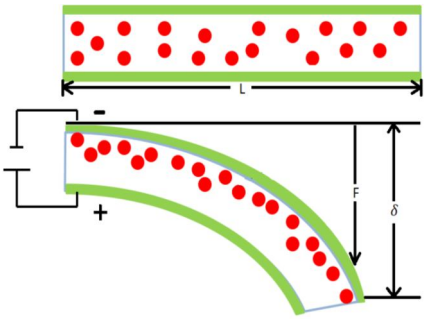

Fig. 4. IPMC deviation

Where L0 is related coefficient to Onsager equations in standard formulation (The coupled coefficient) and E is the Electric Field. So the pressure varies along length x and remains constant along width

$$P(x) = \frac{1}{w} \frac{\partial F(x)}{\partial x} = \frac{L'}{k} \frac{t^3 E}{x^2} \qquad (10)$$

Net compressive pressure

$$P_{comp}(x) = P(x) - P_i \qquad (11)$$

Where,

$$P_i = P_{blood} - P_{atm} \qquad (12)$$

There's another pressure(offered by the stem cell membrane that also plays a role PM(r) at E=0;

$$P_M \sim \frac{8Tension}{2R + L} \sim \frac{8Tension(E)}{2r(E) + L} \qquad (13)$$

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r decreases so does the Tension, hence keeping PM nearly constant Therefore,

$$F_{comp}(x) = w\left(\frac{L'}{k}\frac{t^3E}{x} + (P_M - P_i)\,x\right) \qquad (14)$$

So we have n strips, each of width w, thickness t, length L, young's modulus Y, moment of inertia I, these strips are attached to the thin surface of the stem cell biological sac. Also, they are so tightly packed together, not allowing the membrane to bulge outward and reducing the efficiency of the pumping action. Now the structure made out of stem cell technology is analytically similar to and can be modelled as two ellipsoids sticking together as the chambers of heart as discussed in the section above. Since this bio-membrane is so thin and elastic, it can be considered that it has least contribution towards the bending of strip hence it is safe to consider young's modulus of the strip only and no contribution of the membrane properties in the bending phenomenon except wall pressure. In the same way I (moment of inertia) remains same. Considering a single strip, it has the following bend
equation where x varies along the length of the strip from equations (9) and (14):

$$\delta(x) = \frac{4F_{comp}(x)x^3}{Ywt^3} \qquad (15)$$

$$\delta(x) = \frac{4\left(\frac{L'}{k}\frac{t^3E}{x} + (P_M - P_i)x\right)x^3}{Yt^3} \qquad (16)$$

Now let's consider one chamber, assuming that this Stem cell generated membrane chamber is ellipsoidal in shape, formed by rotating an ellipse along its longer axis such that it has a circle of radius R when cut horizontally, and height L when cut vertically, the centre of the ellipsoid is at the origin of the cylindrical coordinate system (r, z, _) such that the elongation of ellipsoid is aligned with the z axis. So 0 < x < _r i.e. x = r_, the IPMC strip covers the half of the ellipsoid, and the applied electric field is a function of time (T)

$$\delta(r\theta, T) = \frac{4\left(\frac{L'}{k}t^3E(T)(r\theta)^2 + (P_M - P_i)(r\theta)^4\right)}{Yt^3} \qquad (17)$$

$$r = R\sqrt{1 - \frac{4z^2}{L^2}} \qquad (18)$$

These strips are already in uniformly bend position, and manufactured in this manner, so no stray potential exists. Assuming the strips are attached in such a manner that its contact along vertical cross section varies along z direction approximately continuously. Considering only the half section
of model bio-electronic heart, the volume integral becomes:

$$V(T) = 2\int_{z=0}^{z=\frac{L}{2}}\int_{\theta=0}^{\theta=\pi}\int_{r=0}^{r=C} r\,dr\,d\theta\,dz \qquad (19)$$

where,

$$C = R\sqrt{1 - \frac{4z^2}{L^2}} - \delta(z, \theta, T) \qquad (20)$$

$$\begin{aligned} V(T) = {} & \tfrac{1}{3}\pi LR^2 + \left(\tfrac{4(P_M - P_i)}{Yt^3}\right)^2\left(\tfrac{64LR^8\pi^9}{2835}\right) \\ & + \left(\tfrac{4(P_M - P_i)}{Yt^3}\right)\left(\tfrac{LR^5\pi^6}{32}\right) + \left(\tfrac{4L'}{Yk}E(T)\right)^2\left(\tfrac{4LR^4\pi^5}{75}\right) \\ & - \left(\tfrac{4L'}{Yk}E(T)\right)\left(\tfrac{4(P_M - P_i)}{Yt^3}\right)\left(\tfrac{16LR^6\pi^7}{245}\right) \\ & - \left(\tfrac{4L'}{Yk}E(T)\right)\left(\tfrac{LR^3\pi^4}{16}\right) \end{aligned} \qquad (21)$$

Assuming isobaric model with respect to blood pressure (PM - Pi) = 0 and due to equation (13)

$$V(T) = \frac{1}{3}\pi LR^2 + \left(\frac{4L'}{Yk}E(T)\right)^2\left(\frac{4LR^4\pi^5}{75}\right) - \left(\frac{4L'}{Yk}E(T)\right)\left(\frac{LR^3\pi^4}{16}\right) \quad (22)$$

- $Y = 0.5GPa = 0.5 * 10^10g/(cms^2)$,
- $P_i = 100(mmHg) * (101325/76)(g/s^2cm)/mmHg$,
- $k = 2 * 10^-14cm^3s/g$,
- $R = 3cm\ L = 6cm\ t = 0.08cm$.

Now, from the paper [13]

$$L' = \frac{nev}{\zeta} \quad (23)$$

- n is number density of ions $= N/V = \zeta\sigma/e^2$,
- v is Hydrodynamic Volume ,
- ζ is friction coefficient $= \eta_0 a$,
- η_0 viscosity of solvent $(8.90 * 10^-3g/cms)$,
- a is such that $(a/6\pi)$ is hydrodynamic radii (for Na^+ Ion $1.84 * 10^-8cm$).

Now from [15]

$$v = \frac{4\pi[\eta]M}{\mu} \quad (24)$$

M is the molecular weight = 10πNA; NA = 6:0221409e + 23 (Avogadro's constant)
Again from [15]

$$[\eta] = \frac{\sqrt{2}}{c}\left(\frac{\eta}{\eta_o} - 1 - \ln\left(\frac{\eta}{\eta_o}\right)\right)^{1/2} \quad (25)$$

- $[\eta]$ is intrinsic viscosity,
- η viscosity of solution,
- $c = MN/(VN_A)$ is the weight of particle per unit volume,

Combining the above three equations we get,

$$L' = \frac{2\sqrt{2}e}{5\eta_o a}\left(\frac{\eta}{\eta_o} - 1 - \ln\left(\frac{\eta}{\eta_o}\right)\right)^{1/2} \quad (26)$$

some realistic value L' that satisfies both the equations.

$$L' \geq \frac{Yk}{8E_o}\frac{75}{128R\pi} \quad (27)$$

Where Eo is maximum Electric field (equivalently maximum applicable voltage) that doesn't affect the life of the strip and is suitable with respect to response speed. From graph, the required value of L0 _ 0:0000011659 was found using equation (22) to get minima at E = 1, equation (27) is the best way to get L0. Matlab simulation with real values is in Fig. 5. The final equation

$$V_h = \frac{1}{3}\pi LR^2 + V(T) \quad (28)$$
$$V_{max} = \frac{2}{3}\pi LR^2$$

$$E(T) = \frac{Yk}{4L'}\frac{75}{128\pi}\left(1 + \cos\left(H\frac{9.87T}{\pi}\right)\right) \quad (29)$$

Where H is Heart rate 1.17 hz.

$$V(T) = \frac{\pi LR^2}{3} + \left(\left(\frac{75}{128R\pi} + \frac{75\sin\left(1.17\frac{9.87T}{\pi}\right)}{128R\pi}\right)^2 \cdot \left(\frac{(4LR^4\pi^5)}{75}\right)\right) - \left(\left(\frac{75}{128R\pi} + \frac{75\sin\left(1.17\frac{9.87T}{\pi}\right)}{128R\pi}\right) \cdot \left(\frac{(LR^3\pi^4)}{16}\right)\right) \quad (30)$$

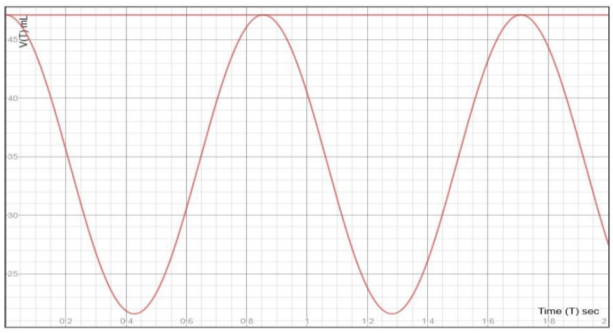

Fig. 5. Plot of volume change in bio-electronic model of Heart

Other analysis such as the restriction to the thickness of IPMC and the response time restrictions and overuse heating will be in further papers, not discussed in the present scope of the model.

V. CONCLUSIONS

The theoretical model has been discussed in mathematical details, results have been established in equations (26) , (27), the model is simulated with the normal heart rate (Fig. 5) which proves and establishes theoretical consistency of the model. Further more by equation (26) , L0 can be calculated
or inversely equation (27) sets the lower bound which will then determine the needed viscosities of the material of IPMC to be used.

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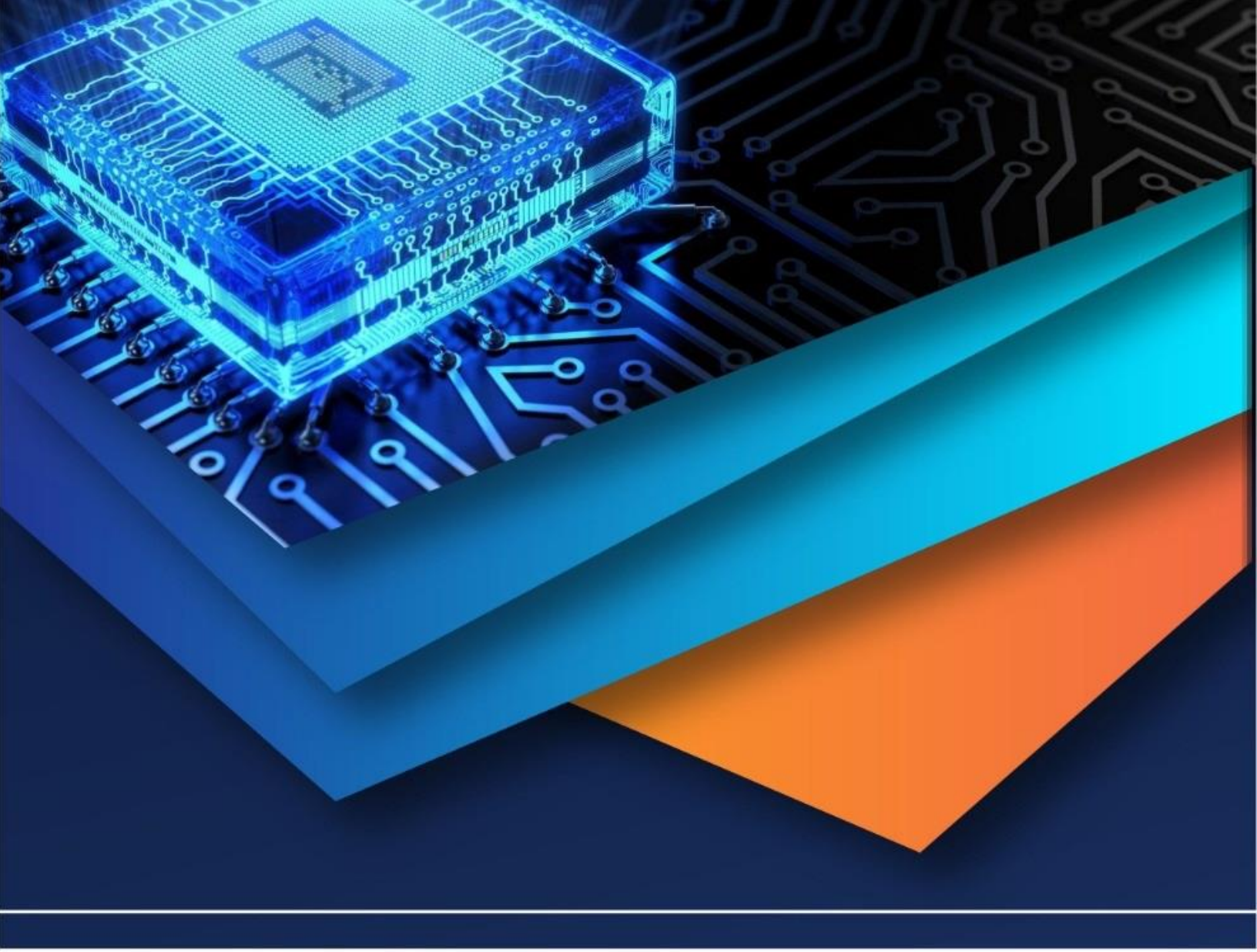

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