Engineering Modeling of Human Cardiovascular System

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Abstract

Cardiovascular system is primarily considered as the human body's transport system. Oxygen, carbon dioxide, nutrients and other vital substances to the various tissues of human body are carried by the blood which circulates in a closed circulation. The cardiovascular system has been comprised of a combination of several basic compartments, which are structurally connected to and functionally interact with each other.

Engineering modeling of such important system has become a useful tool to diagnose the cardiovascular diseases and recommend the appropriate way of their medical treatment. This paper presents a quantified model describing the relationship between the input and output variables of the hemodynamic regulation of the system through implementing a set of first order differential equations that governing this performance and describing its parameters such as pressures, volumes and flows in a closed-loop lumped system. Construction of this model was based on the interaction between the pulsating heart and the vascular circulations of the system through mapping the physiological parameters to the electrical analog elements (resistor, capacitor and inductor and diode) depending upon the close correspondence between the electrical circuits and the cardiovascular system in order to obtain a reasonable investigation for the behavior of the system in normal and pathophysiological conditions.

Keywords *: Cardiovascular System, Lumped Model, Analog Electrical Circuit, Hemodynamic Interaction.*

1. Introduction

The whole cardiovascular system is the body's transport system for respiratory gases, nutritive and waste materials, hormones and heat. It can be approximately divided into seven subsystems including the left and right chambers of the heart, the upper limb and the cerebral circulations, the renal circulation, the splanchnic circulation, the lower limb circulation and the pulmonary circulation, (see the figure 1). Each of these circuits is a combination of arteries, arteriolar and capillary beds, and veins that connect to each other in series. The pulsating heart provides power for the whole circulatory system by rhythmically pumping blood to the peripheral circulation where through arteries, arterioles and capillaries nutrient blood is distributed to organs, tissues and skeletal muscles, then finally is returned back to the right heart chamber by veins. Between the right and left chambers of the heart is the pulmonary circuit where blood is oxygenated. These circuits are not functionally independent but closely associated with each other via concurrent hemodynamic interactions, [1].

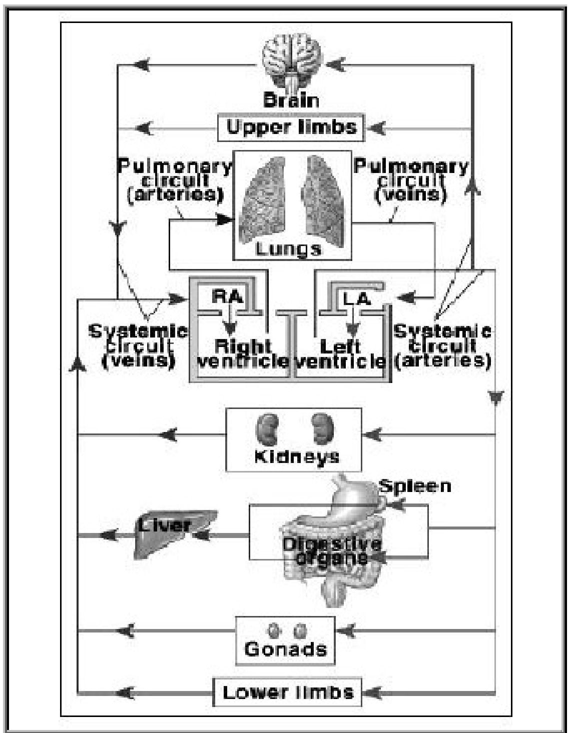

Figure (1) Basic Compartments of the Human Cardiovascular System, [1].

The left ventricle pumps blood into the high pressure areas constituted by the aorta and the arteries of the systemic system. The aorta and its main branches have highly elastic properties. These areas are followed by the arterioles which have less elastic tissue and cause blood pressure to drop. The walls of the arterioles contain more smooth muscle than the walls of the aorta and its primary branches. The blood flow encounters considerable resistance to flow in the arterioles controlled by the degree of muscle constriction which can be significant. This muscular region is followed by the capillaries which contain no muscles. The capillaries are the area where nutrients are exchanged between blood and tissue cells. The level of branching of the vessels reaches its maximum

in the capillaries and the region has a very high surface area, [2].
The capillary vessels discharge their blood into the venules and larger veins. The veins are low pressure vessels and their vessel walls have low elastic properties. The latter makes the veins ideal for the storage of blood. Large volume alterations are archived without significant pressure changes in the veins. The veins contain muscles which allow mobilization of volume to other parts of the cardiovascular system. From the systemic veins the blood is returned to the right heart chamber from which it enters the pulmonary circuit, [2].

The contraction and relaxation of the heart muscles in the heart chamber causes two valves, called the mitral and aortic valve, to open and close due to pressure differences. When modeling the cardiac cycle, one can therefore distinguish four phases.

1. Isovolumic contraction phase, when the mitral and aortic valves are closed. Pressure is being built up in this phase, until the left ventricular pressure rises sufficiently above the aortic pressure to open the aortic valve.
2. Ejection phase, when the mitral valve is closed and the aortic valve is open. Blood flows out of the chamber into the aorta.
3. Isovolumic relaxation phase, when the mitral and aortic valves are closed. Pressure in the chamber decreases until it is so low that the mitral valve opens.
4. Filling phase, when the mitral valve is open and the aortic valve is closed. Blood flows into the chamber, and the cycle then repeats itself.

The first two phases of this cycle are known as systole and the last two phases are known as diastole, [2].

2. Modeling the Cardiovascular System

The cardiovascular system can be described in terms of its hemodynamic variables, i.e. the blood pressure, volume and by the cardiovascular parameters such as compliances and resistances in the corresponding compartments. Different models have been developed to investigate it and various tools, ranging from rather simple to very sophisticated numerical techniques, have been employed. The simplest and most effective way to mathematically represent such a complex system is the lumped parameter modeling method where an electrical analog is frequently employed since a close correspondence between the hemodynamic parameters such as blood pressure, flow and resistance can be well mapped by the corresponding electrical elements including voltage, current, diode, resistor, inductor and capacitor, [3], [4]. In this analogy that we will use, electrical charge represents blood volume, while potential (difference) and currents correspond to pressure (difference) and flow rates. A particular vessel, or group of vessels, can be described by an appropriate combination of resistors, capacitors and inductors. Blood vessels' resistance, depending on the blood viscosity and the vessel diameter, is modeled by resistors. The ability to accumulate and release blood due to elastic deformations, the so-called vessel compliance, is modeled by capacitors. The blood inertia is introduced using coils, and finally heart valves (forcing unidirectional flow) are modeled by diodes, [4].

3. Vessel Resistance and Electrical Resistance

Blood flowing from wider arteries into smaller arterioles encounters a certain resistance. This resistance can be modeled as follows. Consider an ideal segment of a cylindrical vessel. The pressure difference between its two ends and the flow through the vessel depend on each other. Although this dependence will in general be nonlinear, for a laminar flow (which is the type of flow we are interested in) it can be accurately approximated by a linear relation, [5]. If we indicate by Rc the proportionality constant between the pressure difference P and the flow F then it can be written

$$Rc = P / F \tag{1}$$

Similarly, a resistor is an electronic component that resists an electric current by producing a potential difference between its end points. In accordance with Ohm's law, the electrical resistance Re is equals to the potential difference V across the resistor divided by the current I through the resistor

$$Re = V / I \tag{2}$$

4. Vessel Compliance and Capacitance

The walls of the vessels are surrounded by muscles that can change the volume and pressure in the vessel. Consider the blood flow into such an elastic (compliant) vessel. We denote the flow into the vessel by Fi and the flow out of the vessel by Fo. Then the difference F = Fi − Fo which corresponds to the rate of change of blood volume in the vessel is related to a change of pressure P inside the vessel. Assuming a linear relation, it has been that

$$F = Cc\, dP/dt \tag{3}$$

where Cc is a constant related to the compliance of the vessel.
The analogy with a capacitor is immediate. A capacitor is an electrical device that can store energy

between a pair of closely-spaced conductors, so-called "plates". When a potential difference is applied to the capacitor, electrical charges of equal magnitude but opposite polarity build up on each plate. This process causes an electrical field to develop between the plates of the capacitor which gives rise to a growing potential difference across the plates. This potential difference V is directly proportional to the amount of separated charge Q (e.g. Q = Ce V). Since the current I through the capacitor is the rate at which the charge Q is forced onto the capacitor (e.g. I = dQ/dt), this can be expressed mathematically as

$I = Ce\ dV/dt$	**4**

where the constant Ce is the electrical capacitance of the capacitor, [5].

5. Blood Inertia and Inductance

Since blood is inert, it follows that when a pressure difference is applied between the two ends of a long vessel that is filled with blood, the mass of the blood resists the tendency to move due to the pressure difference. Once more assuming a linear relation between the change of the blood flow (dF/dt) and the pressure difference P that can be expressed mathematically as

$P = Lc\ dF/dt$	**5**

Note that this is the hydraulic equivalent of Newton's law, which relates forces to acceleration. The inertia of blood can be modeled by a coil (also known as an "inductor"), since the current in a coil cannot change instantaneously, [5]. This effect causes the relationship between the potential difference V across a coil with inductance Le and the current I passing through it, which can be modeled by the differential equation

$V = Le\ dI/dt$	**6**

6. Valves and Diodes

An ideal valve forces the blood to flow in only one direction. More specifically, it always stops the flow in one direction while it allows the blood to flow in the other direction, opposing only a small resistance Rc to the flow, as soon as the pressure difference is higher than a certain critical pressure P* which is often taken to be zero. For this reason it is common use to model the action of a valve as follows:

$F = 0$ if $P < P^*$ $= P/Rc$ if $P \geq P^*$	**7**

The electrical analog of a valve is a diode. In electronic circuits, a diode is a component that allows an electric current I to flow in one direction, but blocks it in the opposite direction. Thus, can use the idealized relationship corresponding to equation (7)

$I = 0$ if $V < V^*$ $= V/Re$ if $V \geq V^*$	8

One advantage of modeling the cardiovascular system by an electrical circuit is that Kirchoff's laws for currents and potential differences can be applied:

- The sum of currents entering any junction is equal to the sum of currents leaving that junction (conservation of blood mass).
- The sum of all the voltages around a loop is equal to zero (pressure is a potential difference).

All these elements, or their nonlinear extensions, are used in different forms in the models for the heart and its environment, the arterial system that can employed in a relatively simple models for the cardiovascular system that is known as Windkessel model, [6].

7. The Windkessel Model

The Windkessel model is based on physical characteristics of vessels and describes the pulse-wave propagation of the blood. Vessels that consist of a relatively large proportion of elastic fibres, such as the aorta, the pulmonary arteries and the adjacent parts of the great arteries, are called windkessel vessels. With each pulse any given segment of such a vessel distends to store a volume of blood. As it subsequently contracts back to its original dimensions, it pushes blood on to the next segment. The name windkessel (German meaning for air chamber) has been given to these vessels and their function because of the resemblance to the air-filled chambers that similarly affect the velocity and pressure of fluids driven by pistons through systems of pipes, [7].
This consists of a capacitor representing the compliant aorta, and a resistor representing the stiffer peripheral vessels, connected in parallel. Earlier variants of this model attempted ordinary differential equations that relate the dynamics of aortic pressure and blood flow to various parameters such as arterial compliance, resistance to blood flow and the inertia of blood to describe either the wave motion of the blood or the regulatory processes, but dealt separately with individual sections of the cardiovascular system, [7].

8. Basis of the Engineering Model

The proposed model constitutes a generalization of the mathematical model described by Ursino 1998, [8]. The main improvement is considering the heart as a pulsatile pump, therefore, the activity of the ventricles may be simulated by an elastance variable model. Also, the vascular system is sub-divided into eight compartments, as shown in the figure (2). Five of these compartments are used to reproduce the systemic circulation, differentiating among the systemic arteries (subscript sa), the splanchnic peripheral and venous circulations (subscripts sp and sv, respectively), and the extrasplanchnic peripheral and venous circulations (subscripts ep and ev, respectively). Similarly, the other three compartments mimic the arterial, peripheral and venous pulmonary circulations (subscripts pa, pp, and pv, respectively). Subscripts la, lv, ra, and rv indicate the left atrium, left ventricle, right atrium, and right ventricle, respectively. Each compartment includes a hydraulic resistance Rj, which accounts for pressure energy losses in the jth compartment, a compliance Cj, which describes the amount of stressed blood volume stored at a given pressure P, flow F and an unstressed volume Vu,j. For the sake of simplicity, the inertial effects of blood have been included only in the large artery compartments, in which blood acceleration is significant, [8],[9].

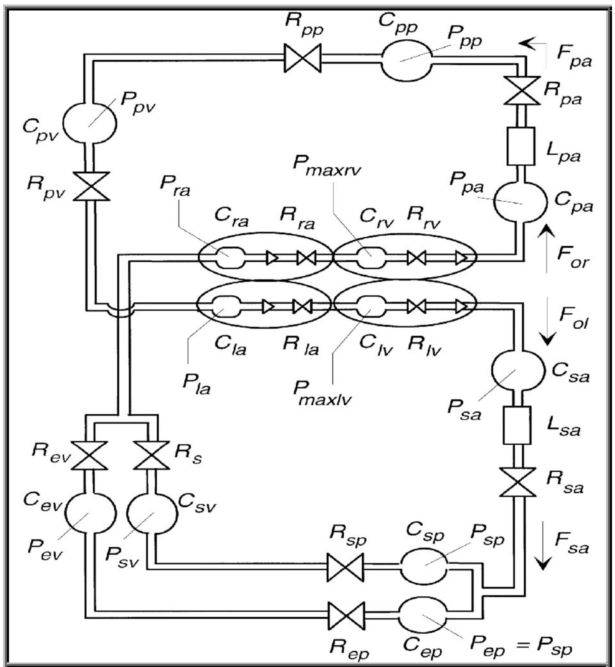

Figure (2) : Hydraulic analog of the cardiovascular system.

Equations relating pressure and flow in all points of the vascular system have been written by enforcing preservation of mass at the capacities in figure (2) and equilibrium of forces at the inertances Lj and by assuming that the total amount of blood initially contained in the vascular system is 5,300 ml, [10].

A nonlinear closed-loop lumped parameter model of the intact pulsatile heart and its circulations is illustrated in figure (3). The models for the right and the left heart are similar, with different values of parameters. The atrium is described as a linear capacity characterized by constant values of compliance and unstressed volume, i.e., the contractile activity of the atrium is neglected. Blood passes from the atrium to the ventricle through the atrioventricular valve, mimicked as the series arrangement of an ideal unidirectional valve in series with a constant resistance. Also, a distinction is made between the capacitance, inertia and other characteristics in the different components of the system. For example; the capacitance and resistance for blood flow obviously depend on the width of the arteries and veins, which may vary considerably within the human body. Notice that the time-varying pressures generated by the heart muscles in the left and right ventricles are indicated by capacitors with arrows through them, while we have also used the standard symbol for electrical earth to indicate a point of zero voltage, which corresponds to a reference pressure based on which all other pressure differences are stated.

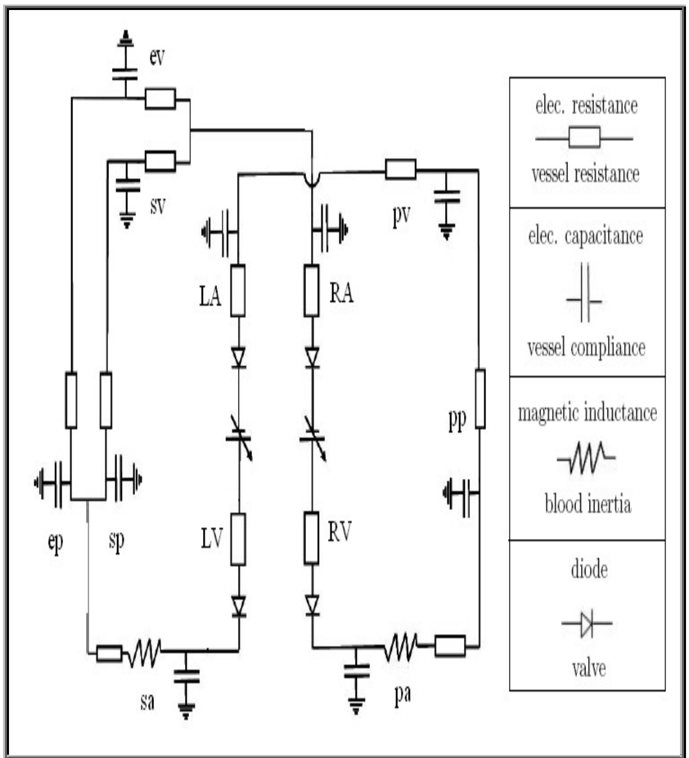

Figure (3) A closed-loop lumped parameter model of the cardiovascular system.

The model for pressures, volumes and flows then becomes as follows.
Conservation of mass and balance of forces in the different compartments lead to

$$\frac{dP_{pp}}{dt}=\frac{1}{C_{pp}}\left(F_{pa}-\frac{P_{pp}-P_{pv}}{R_{pp}}\right)$$
$$\frac{dP_{pv}}{dt}=\frac{1}{C_{pv}}\left(\frac{P_{pp}-P_{pv}}{R_{pp}}-\frac{P_{pv}-P_{pl}}{R_{pv}}\right) \quad (9)$$

$$\frac{dP_{pa}}{dt}=\frac{1}{C_{pa}}(F_{o,r}-F_{pa})$$
$$\frac{dF_{pa}}{dt}=\frac{1}{L_{pa}}(P_{pa}-P_{pp}-R_{pa}F_{pa}) \qquad (10)$$

For the pulmonary arteries in the upper circuit of the figure, while it leads to the following for the lower circuit of the model.

$$\frac{dP_{sp}}{dt}=\frac{1}{C_{sa}}(F_{o,l}-F_{sa})$$
$$\frac{dF_{sa}}{dt}=\frac{1}{L_{sa}}(P_{sa}-P_{sp}-R_{sa}F_{sa})$$
$$\frac{dP_{sp}}{dt}=\frac{1}{C_{sp}+C_{ep}}(F_{sa}-\frac{P_{sp}-P_{sv}}{R_{sp}}-\frac{P_{sp}-P_{ev}}{R_{ep}})$$
$$\frac{dP_{ev}}{dt}=\frac{1}{C_{ev}}(\frac{P_{sp}-P_{ev}}{R_{ep}}-\frac{P_{ev}-P_{ra}}{R_{ev}}) \qquad (11)$$

Finally, for the left and right atria

$$\frac{dP_{la}}{dt}=\frac{1}{C_{la}}(\frac{P_{pv}-P_{la}}{R_{pv}}-F_{i,l})$$
$$\frac{dP_{ra}}{dt}=\frac{1}{C_{ra}}(\frac{P_{sv}-P_{ra}}{R_{sv}}+\frac{P_{ev}-P_{ra}}{R_{ep}}-F_{i,r}) \qquad (12)$$

Here, Fi,l and Fo,l are the flow into and out of the left ventricle, and Fi,r and Fo,r are the flow into and out of the right ventricle (in ml/s). Assuming a known and constant total blood volume Vo (5,300 ml), it can express the last remaining pressure, the splanchnic venous pressure Psv, in terms of all other pressures

$$P_{sv}=\frac{1}{C_{sv}}\{V_o-C_{sa}P_{sa}-(C_{sp}+C_{ep})P_{sp}-C_{ev}P_{ev}-C_{ra}P_{ra}$$
$$-V_{rv}-C_{pa}P_{pa}-C_{pp}P_{pp}-C_{pv}P_{cp}-C_{la}P_{la}-V_{lu}-V_u\} \qquad (13)$$

The left and right ventricles are modeled using state variables that represent volumes instead of pressures and the flows through the valves

$$\frac{dV_{rv}}{dt}=F_{i,r}-F_{o,r}$$
$$F_{i,r}=0 \quad if \quad P_{ra}\le P_{rv}$$
$$=\frac{P_{ra}-P_{rv}}{R_{ra}} \quad if \quad P_{ra}>P_{rv}$$
$$F_{o,r}=0 \quad if \quad P_{max,rv}\le P_{pa}$$
$$=\frac{P_{max,rv}-P_{pa}}{R_{rv}} \quad if \quad P_{max,rv}>P_{pa}$$
$$\frac{dV_{lv}}{dt}=F_{i,l}-F_{o,l}$$
$$F_{i,l}=0 \quad if \quad P_{la}\le P_{lv}$$
$$=\frac{P_{la}-P_{lv}}{R_{la}} \quad if \quad P_{la}>P_{lv}$$
$$F_{o,l}=0 \quad if \quad P_{max,lv}\le P_{sa}$$
$$=\frac{P_{max,lv}-P_{sa}}{R_{lv}} \quad if \quad P_{max,lv}>P_{sa} \qquad (14)$$

where Pmax,rv and Pmax,lv, are the right and left ventricle pressures in isometric conditions respectively and the pressures and resistance in these ventricles are given by

$$R_{lv}=k_{r,lv}P_{max,lv}$$
$$P_{lv}=P_{max,lv}-R_{lv}F_{o,l}$$
$$R_{rv}=k_{r,rv}P_{max,rv}$$
$$P_{rv}=P_{max,rv}-R_{rv}F_{o,r}$$
$$P_{max,lv}(t)=\phi(t)E_{max}(V_{lv}-V_{u,lv})+\{1-\phi(t)\}P_{o,lv}[Exp(k_{E,lv}V_{lv})-1]$$
$$P_{max,rv}(t)=\phi(t)E_{max}(V_{rv}-V_{u,rv})+\{1-\phi(t)\}P_{o,rv}[Exp(k_{E,rv}V_{rv})-1] \qquad (15)$$

At diastole, when the cardiac muscle fibers are relaxed, the ventricle fills through an exponential pressure/volume function, which reflects the elasticity both of the relaxed muscle and of its external constraints (mainly the pericardium). In contrast, a linear pressure/volume function at end systole whose slope is adopted (usually called the ventricle elastance at the instant of maximal contraction,) is denoted by Emax. The constants kr and kE describe the ventricle resistance and the end-diastolic pressure-volume relationship for the heart and Vu is the unstressed ventricle volume.

Shifting from the end-diastolic to the end-systolic relationship is governed by a pulsating activation function Φ(t), with period T equal to the basal cardiac cycle (= 0.833 sec), that steers Pmax,lv, the isometric left ventricle pressure. This function is controlled by the baroreflex control system, which is a highly complex function of the sinus nerves. For simplicity, we approximate the ventricle activation function by a simple sine function with ω = 1.25 rad, the signal frequency which corresponds to the cardiac cycle.

$$\phi(t) = \sin(2\pi\omega) \qquad 0 \le \sin(2\pi\omega)$$
$$\quad = \quad 0 \qquad \sin(2\pi\omega) < 0 \qquad (16)$$

Finally, blood flow leaving the ventricle depends on the aortic valve opening and on the difference between the isometric ventricle pressure and arterial pressure (afterload). The simulated time patterns of some hemodynamic quantities (left ventricle pressure-volume function, systemic arterial pressure, left ventricle pressure, cardiac output) during basal cardiac cycle are shown in the figure (4), [11].

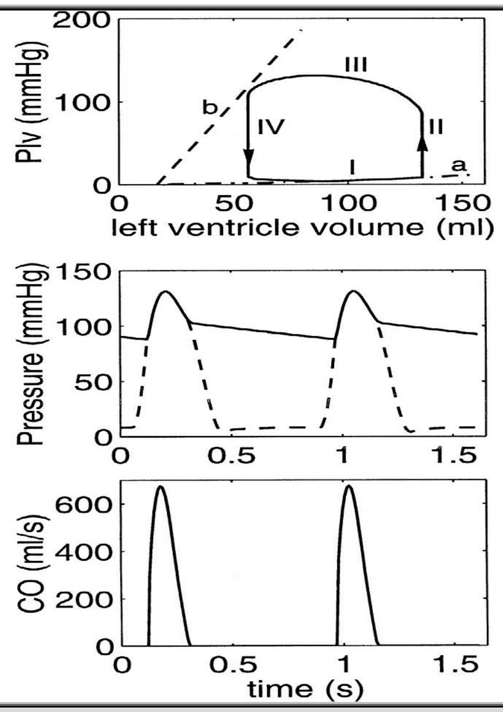

Figure (4) Left ventricle pressure (Plv)/volume function during a basal cardiac cycle (top). Dotted lines marked a and b represent diastolic and end-systolic pressure/volume functions, respectively, in isometric conditions: I, filling phase; II, isometric contraction; III, ejection phase; IV, isometric relaxation. Middle: time pattern of systemic arterial pressure (continuous line) and Plv (dashed line). Bottom: time pattern of cardiac output (CO) from the left ventricle.

For the present, however, instead of solving the equations directly, we have obtained a solution starting from the known physical and physiological properties of the cardiovascular system. The effects of all experimentally observed processes on one pumping cycle of the heart were considered and a mathematical description of oscillations in the blood flow was acquired. The strategy that used to achieve this goal may be comprised into two steps:

(1) The desired average pressure and volume distribution and cardiac output determine the parameter values of compliances and resistances.

(2) Inductances and elastance functions are subsequently assigned values that produce representative pressure and flow pulses.

The selection of cardiovascular parameter values in the human circulation model is guided by data suitably rescaled for a subject with a 70 kg body weight. The goal is to obtain realistic average pressure levels and volume distribution in the system. In addition, computed ventricular pressure, root aortic pressure and outflow curves should closely resemble the corresponding human ones, [11]. The parameter values characterizing the cardiovascular system in basal condition are illustrated per the following table.1;

Table (1) Parameter values characterizing the cardiovascular system in basal condition.

Compliance (ml/mmHg)	Unstressed Volume (ml)	Hydraulic Resistance (mmHg . sec/ml)	Inertance (mmHg . ml/sec^2)
C_{sa} = 0.28	$V_{u,sa}$ = 0	R_{sa} = 0.06	L_{sa} = 0.22 x 10^{-3}
C_{sp} = 2.05	$V_{u,sp}$ = 274.4	R_{sp} = 3.307	
C_{ep} = 1.67	$V_{u,ep}$ = 336.6	R_{ep} = 1.407	
C_{sv} = 61.11	$V_{u,sv}$ = 1,121	R_{sv} = 0.038	
C_{ev} = 50.0	$V_{u,ev}$ = 1,375	R_{ev} = 0.016	L_{pa} = 0.18 x 10^{-3}
C_{pa} = 0.76	$V_{u,pa}$ = 0	R_{pa} = 0.023	
C_{pp} = 5.80	$V_{u,pp}$ = 123	R_{pp} = 0.0894	
C_{pv} = 25.37	$V_{u,pv}$ = 120	R_{pv} = 0.0056	

While the parameters describing the four right and left chambers of the heart are illustrated per the following table.2;

Left Heart	**Right Heart**
C_{la} = 19.23 ml/mmHg	C_{ra} = 31.25 ml/mmHg
$V_{u,la}$ = 25 ml	$V_{u,ra}$ = 25 ml
R_{la} = 2.5 x 10^{-3} mmHg.sec/ml	R_{ra} = 2.5 x 10^{-3} mmHg.sec/ml
$P_{o,lv}$ = 1.5 mmHg	$P_{o,rv}$ = 1.5 mmHg
$k_{E,lv}$ = 0.014 / ml	$k_{E,rv}$ = 0.011 / ml
$V_{u,lv}$ = 16.77 ml	$V_{u,rv}$ = 40.8 ml
$E_{max,lv}$ = 2.95 mmHg/ml	$E_{max,rv}$ = 1.75 mmHg/ml
$k_{R,lv}$ = 0.375 x 10^{-3} sec/ml	$k_{R,rv}$ = 1.4 x 10^{-3} sec/ml

9. Discussion

Engineering models of the cardiovascular system are useful for a deeper understanding of complex process occurring in the heart and blood vessels in the normal and pathophysiological conditions. All the findings that were done from derived pressure-volume relations or evaluated hemodynamic variables may serve as a useful assistant tool for diagnosing a patient illness and then for recommending of the way of medical treatment of the cardiovascular disorders.

The human cardiovascular system can be described in terms of quite complex fluid dynamics. Due to this highly complicated structure of the system, high dimensional numerical approaches are always limited to studies on local hemodynamics. In contrast, lumped parameter models, which mathematically represents the characteristics of hemodynamics with a series of relationships among pressure, volume and flow in various compartments of the system, may provide a quantitative insight into the behavior of the whole system. Construction of this model was implemented based on a phenomenological characterization of hemodynamics using an electrical analog circuit.

10. Conclusion

The proposed model for the human cardiovascular system has adopted a nonlinear lumped parameters model of the intact pulsatile heart and its vascular circulations by using an electrical analog circuit which closely corresponds to the system. This means a distinction is made between veins, which carry blood to the heart, and the arteries, which take blood from the heart to the organs, and between the pulmonary system, which corresponds to the lungs, the splanchnic system, which corresponds to abdominal internal organs, the peripheral system, corresponding to the outer part of the body, and the extrasplanchnic system, which corresponds to other organs. Inasmuch the human cardiovascular system is a closed-loop system, hemodynamics in any separate segment is determined by hemodynamic interactions through out the whole system.

The structural pattern of this cardiovascular model follows the architecture of many previous cardiovascular descriptions and may only differ by the number of sections used to characterize the system. The desired average pressure and volume distribution and cardiac output determine the parameter values of compliances and resistances. Most of the hemodynamic patterns presented by the model were consonant with the clinical measurements.

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نمذجة هندسية للجهاز القلبي الوعائي في جسم الإنسان

حسنين علي لفته موسى

قسم الهندسة الطبية / كلية الهندسة / جامعة النهرين

الخلاصة

يعتبر الجهاز القلبي الوعائي المنظومة الأساسية لنقل الأوكسجين ، ثنائي أوكسيد الكاربون ، الغذاء وغيرها من المواد الحيوية الأخرى إلى كافة أنسجة جسم الإنسان عن طريق دوران الدم خلال دورة مغلقة. من الممكن جداً تقسيم هذا الجهاز إلى عدة مركبات فرعية ترتبط مع بعضها البعض تشريحياً ووظيفياً. لقد أضحت النمذجة الهندسية لمثل هذه الأنظمة الفسلجية المهمة وسيلة مفيدة لتشخيص الأمراض التي تصيب هذا الجهاز مما سيسهل بدوره في تحديد الطريقة الملائمة لعلاجها.

يُقدم هذا البحث أنموذج كمي لتوصيف العلاقة ما بين متغييرات الإدخال والإخراج الخاصة بتنظيم أداء ديناميكية دوران الدم عبر مكونات الجهاز القلبي الوعائي في جسم الإنسان وذلك من خلال تشكيل مجموعة من المعادلات الرياضية التفاضلية ذات الدرجة الأولى التي تحكم هذه الآلية الحيوية وبالإعتماد على ميكانيكية التغذية المغلقة. إن تنفيذ هذا الأنموذج أعتمد وبشكل أساسي على إمكانية التنسيق ما بين القلب النابض ومنظومة الأوعية الدموية للجهاز من خلال التوافق ما بين العوامل الفسلجية الخاصة بتلك الأوعية الدموية (كالحجم والضغط ومعدل الجريان) مع العناصر الأساسية للدوائر الكهربائية الضمنية (كالمقاومة والمتسعة والمحاثة والدايود) بهدف الحصول على نتائج مقبولة ومنطقية قدر الإمكان والتي ستحدد بدورها الأداء الصحيح للجهاز القلبي الوعائي في حالته الطبيعية أو تشخص الإضطرابات المرضية التي تصيب هذا الجهاز الحيوي.