2. Жалдак М.І. Чисельні методи математики : посіб. для самоосвіти вчителів / М.І. Жалдак, Ю.С. Римський. – К., 1984. – 206 с.
3. Применение вычислительных методов в научно-технических исследованиях : межвуз. сб. науч. трудов. – Пенза, 1984. – 160 с.

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ОСНОВЫ ОБЕСПЕЧЕНИЯ ПРОФЕССИОНАЛЬНОЙ КОМПЕТЕНТНОСТИ УЧИТЕЛЯ ФИЗИКИ В ПРОВЕДЕНИИ ЧИСЛЕННЫХ ЭКСПЕРИМЕНТОВ И ПРОЦЕСС ИХ ФОРМИРОВАНИЯ

Рассмотрены методические особенности формирования профессиональной компетентности учителя физики в процессе обработки многочисленных экспериментальных данных. Приведены рекомендации по вычислению студентами абсолютной и относительной погрешностей, выполнении арифметических операций над погрешностями, в частности, операции деления.

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METHODOLOGICAL FOUNDATIONS OF PROFESSIONAL COMPETENCE PHYSICS TEACHER IN NUMERICAL EXPERIMENTS AND PROCESS OF THEIR FORMATION

Methodical peculiarities of professional competence of teachers of physics in the numerical treatment of experimental data. The recommendations on student computing absolute and relative errors, performing arithmetic operations on errors, including division operation.

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DESIGN AND DEVELOPMENT FOR A SIMULATOR OF BIOPHYSICAL CARDIOVASCULAR PROCESSES

This article about design and development for a simulator of biophysical cardiovascular processes. This is necessary to understand different cardiovascular pathologies and the diagnostic methods findings.

Key words: simulator, cardiovascular processes, learning tools.

Abstract. Teaching students the Cardiovascular System and the diverse biophysical factors involved in cardiac pathologies can be greatly supported by simulators. Simulators allow students to explore each variable in the study when they are modified. It is of vital importance to study the cardiovascular system during the formation of medical students and students of related areas. Basically, the cardiovascular system shows heterogeneous biophysical properties that force researchers to design specific simulators for a particular area or segment of the system. For example, the large blood vessels emerging from the left ventricle have elastic properties, while the rest of the vessels irrigating most organs and muscles lose elasticity to become muscle type vessels. We designed and developed a simulator that allows the study of blood flow, that is, applied hydrodynamics to describe blood flow. This is necessary to understand different cardiovascular pathologies and the diagnostic methods findings.

In this sense, the physical principles governing peripheral circulation are not necessarily applied to the study of the isolated functioning of the heart. Experiments in Biophysics laboratories reduce the studies to ideal conditions, far from the real systems. Hence the study of hemodynamics is permanently improving.

For the simulator design, we develop mathematical models of physical processes like flow, systolic and diastolic pressures of the cardiac cycle. This is represented by an electric circuit described by multiparametric differential equations, which can be relating pressure and flow variables.

With this model a simulator of biophysical cardiac processes is developed, solving the differential equations by Euler's method where the left ventricle, aortic valve and the principal arterial ramification are represented by the circuit components.

Furthermore, the behavior of both, the ventricular pressure and the proximal arterial pressure in the aorta can be followed in artery stenosis, using the previous mathematical model. For stenosis simulation, valve resistance is varied, assuming a constant and uniform blood flow, similar to a sinusoidal wave.

On the other hand, regulated blood flow is simulated, based on Windkessel's model. This model is used to describe basic properties of the vascular system and to study the relationship between the hemodynamic variables in large vessels. For flow simulation, cardiac frequency is varied.

Finally blood flow rates velocities are modeled through Poiseuille's model, considering a stable blood flow, viscosity and Navier Stokes equation in newtonian fluid, for a blood vessel whose narrowest segment is much longer than its diameter, and is assumed constant. This model works for blood vessels with small diameter and rigid walls. The simulators were built using Visual Basic 6.0 version, which can be used in any PC compatible computer, Windows OS and 40.4 MB available space on hard drive.

1. INTRODUCTION

Universities and research centres have been recently involved in the development of computer models and simulation of diverse phenomena aiming for research tools and development. Participation of multidisciplinary groups competent in different research areas are needed to build these computer models. As a result, different simulators of complex phenomena are available on line. With this in mind, a simulator of several biophysical cardiovascular processes is here described.

To understand the mechanisms of irrigation in tissues and the delivery of nutrients and oxigen to the cells in the human body, as well as CO_2 and waste elimination, it is necessary to study cardiovascular physiology and pathology. In the cardiovascular system, blood flows by pressure, from high to low, due to cardiac action. It flows in one direction because the valves redirect blood flow.

For blood flow to be continuous, large arteries need elasticity. Arteries also need more muscular content and less flexibility for blood to flow preferentially toward tissues. Therefore variations in arterial diameter significantly modify peripheral resistance. The presence of blood capillary vessels with small diameters diminishes blood flow, and their thin walls allow metabolic gas exchange (1). It can be observed that this is a hemodynamic heterogeneous system.

There is no computer yet capable to process each and every function of the cardiovascular system simultaneously. However, it is possible to reduce or divide the system by simulation strategies.

In this study, the cardiovascular system is reduced through its equivalent electric circuit.

In this case the resistance to blood flow by the diameter of thousands of peripheral blood vessels can be represented by a single electric resistance; the increase in value means an increase in arterial pressure. The minimal pressure necessary to keep a continuous flow in the closed system can be electrically represented by a battery [2; 3].

The next section of our simulator is the study of blood flow.

Our body contains approximately 5 liters of blood, pumped by the heart to flow through the blood vessels, travelling 120 times the whole body every hour. To simulate this, we use Windkessel's Theory, proposed by the german physiologist Otto Frank in 1899. This theory takes arteries elasticity into account and describes how the heart pumps a volume of blood during systole, part of which is stored in the arteries because the pressure causes a radial extension of the arteries.

Being flexible, arteries deform increasing in diameter. During diastole, the heart is not pumping blood, but the stored blood flows, because arteries contract through the elastic energy stored during systole, allowing the continuity of the flow. Therefore blood flow during systole is sinusoidal, while no flow is pumped during diastole. If arteries were rigid, the result would be an interrupted and discontinuous flow.

Now, assuming flexible arteries, as it actually is, their diameter increase during systolic increase in pressure, storing (trapping?) part of the blood flow. Later, during diastole, the trapped (stored) blood is released. The result is a continuous and basically uniform flow. Windkessel's theory gives an adequate simplification to obtain a simple global model of the cardiovascular system at the artery level [6; 7; 8].

Viscous fluids in hydrodynamics are described by Navier-Stokes equation. Therefore it is possible to apply this equation in hemodynamics when blood density, pressure difference (proximal and distal), and viscosity are considered. Density factor introduces a non-linear component. In normal conditions in a person, this component can be disregarded, so the equation is reduced to Stokes equation, assuming a non-compressible fluid.

The solution to this equation brings us to Poiseuille law, which applies at the narrowest part of blood vessels, where the length is much longer than the diameter length and the vessel is considered rigid, uniform and straight.

For the first version in the design of the simulator, the following is considered:

1) the vessel is rigid at the stenosis region
2) the radial flow component is negligible, since only average flow and velocities are taken into account
3) blood density changes are negligible under these conditions.

2. SIMULATOR MODELS

The model used for the simulator of the basic cardiovascular physiology corresponds to an equivalent electric circuit (*fig. 1*). The cardiovascular system is modelled with a capacitor representing the left ventricle. The aortic valve is represented by a diode and a resistance; while the aorta is modelled by an inductance, its flexibility corresponding to two capacitors, a proximal and a distal one.

All peripheral vessels are represented by a unique resistance. In order to maintain an initial pressure a battery is necessary. This circuit has three nodes where the electric equivalent of the pressures: ventricular and aortic (proximal and distal) is registered (2). The electric circuit models the circulatory system, from the left ventricle to the return veins. It starts with the maximum pressure registered during the left ventricular systole. At arterioles, pressure decreases rapidly as they become more distal. Oscillation stops because the vessels walls are more rigid at this point.

Pressure at capillaries keeps decreasing up to a minimum at the veins. Right away the pressure rises due to the right ventricle systole and the increase in blood pressure at the lungs, caused by the right ventricle ejection. The cycle repeats itself. To produce circuit oscillations, elastance (the inverse of capacitance) is considered (2, 3). Elastance variations should faithfully follow ventricluar systole time. As a result, elastance is variable.
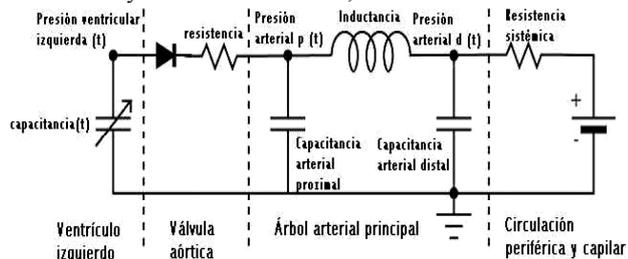

Fig. 1. Circuit representing the cardiovascular system

The contraction-relaxation function of the left ventricle consists on changing the left ventricle capacity, simulating (emulating) the ventricular contraction and relaxation.

The elastance function is implemented using a polynomial representation by sections, in the following way:

$$Elv(t) = e_{min} + ct^2 + c2t^3 + c3t^{4`} \qquad (1)$$

for $t < tc$.

$$Elv(t) = e_{max} + r\,([t - tc])^2 + r^2\,(t - tc)^3 + r3(t - tc)^4 \qquad (2)$$

for $t > tc$.

To simulate blood flow it has to take into account the periodic behavior of the heart. In fact, the blood flows through the arteries very similar to a sinoidal wave, where the amplitude is the maximum and the frequency is the heart rate. It is described by a differential equation called the wave equation:

$$\frac{\partial^2 Q(x,t)}{\partial t^2} - c^2 \frac{\partial^2 Q(x,t)}{\partial x^2} = 0 \; . \qquad (3)$$

This function refers to the amount of blood passing through one point in the artery at a given time. This equation implies the second derivative in time (*t*) and the second derivative in space (*x*), suggesting that the solution should be similar in time and space, where «c» is the rate of propagation of the wave. What matters now is to find the solutions to the wave equation.

Using separation of variables, it is reduced to:

$$\frac{\partial^2 f(t)}{\partial t^2} = Dc^2 f(t) \; . \qquad (4)$$

$$\frac{\partial^2 g(t)}{\partial x^2} = -Dg(t) \; . \qquad (5)$$

Obtaining

$$g(x) = A\,sen(\sqrt{D}x) = Bcos(\sqrt{D}x) \; . \qquad (6)$$
$$f(x) = A\,sen(c\sqrt{D}t) = Bcos(c\sqrt{D}t) \; . \qquad (7)$$

Leading to the solution

$$Q(t) = A\,sen(\frac{2FC2\pi}{60}t) + B \; . \qquad (8)$$

Where FC: cardiac frequency

A: difference between max minus min flow

B: minimal flow

for $FC \geq 150$.

Lastly for the flow and velocity simulator Poiseuille is used, corresponding to the narrowest part of blood vessels. Considering blood in a Newtonian fluid, viscosity does not change in time or space during one cardiac cycle. However this viscosity could change from patient to patient or in pathology cases. This is because viscosity depends on the amount of red blood cells, hematocrite and temperature.

Furthermore a model of a rigid and cylindrical vessel is considered. Peripheral resistance is controlled by the arterioles. Since no pulse nor oscillation is found here, they can be considered rigid, their elasticity is negligible and the length of the tube is larger than its diameter. Finally a stable flow blood flow is assumed, neither pulsed nor turbulent, in which case Navier Stokes equation would be sed.

Negative sign indicates that blood is moving to the periphery.

Due to viscosity, blood flow rate is grater at the central part of the vessel than at the walls. These are the mathematical models used for the Simulator design.

3. RESULTS

In this version of the simulator several modules are developed.

After installation, the start screen shows up. The menu allows access to each simulator through links. There are three main simulators: blood flow simulator, Poiseuille Law with forks in the simulator, and Basic Cardiovascular Physiology, which is divided into: elastance and velocity, normal conditions, aortic stenosis and hypertension, which depict some pathologies. The first button applies only to the aorta, an elastic artery. It is the blood flow simulator, for which we use Windkessel's law. The program was designed to simulate hemodynamic aspects of large blood vessels, i. e. the aorta under normal conditions. The heart is considered a pump ejecting blood during systole. During diastole, blood flow is continuous due to the elasticity of the aorta (*fig 2*).

For a healthy adult patient the following is considered:

Cardiac frequency (FC) is 72 beats/min

Blood flow at rest is 5 L/min

Flow rate is 0.083L/s

The user interface is shown in *fig. 1*. The middle window shows the oscilloscope plotting blood flow vs. time. To the right there are two boxes, the upper frame is the calculated maximal flow, the lower one shows the minimal flow. At the upper right an animation of the heart is shown.
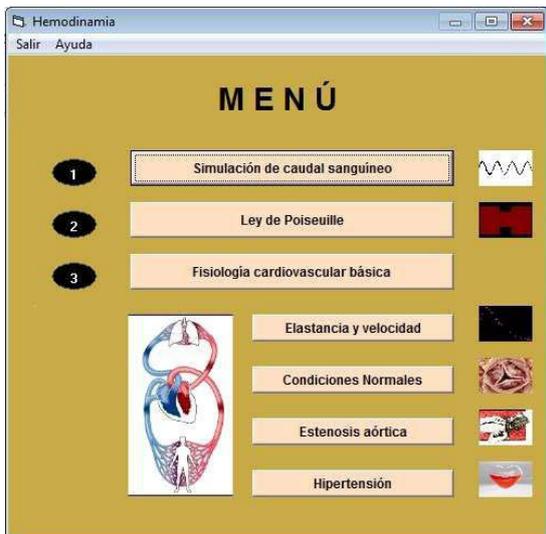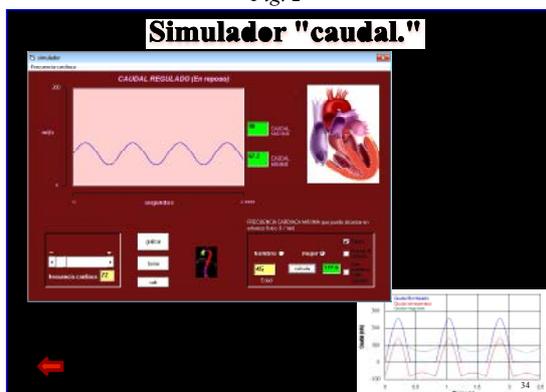
Fig. 2

Fig. 3. *Blood flow Simulator*

Located at the lower central part you find three buttons: 1) plot, 2) erase and 3) exit. The user starts the simulation by pushing the plot button. Input could be age and cardiac frequency. The simulator calculates maximum cardiac frequency.

Varying cardiac frequency you can simulate conditions for rest or exercise. *Fig 2* shows three simulations: 1) at rest, 2) bradycardia and 3) tachycardia. Maximum and minimum flow change according to these conditions; it rises during exercise and decreases with bradycardia.
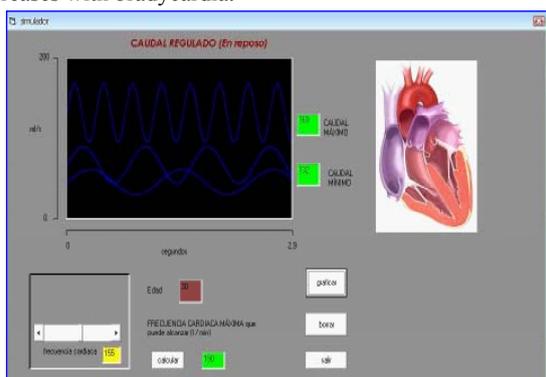

Fig. 4. *Regulated flow Simulator*

For Poiseuille law the following is considered: pressure change, the radius of the blood vessel and the length of the arteriole segment. The start screen shows the first two scrollbars for the pressure corresponding to small arteries, with values in KPa and Hgmm. The next bar modifies the value of the arteriole radius and length, the latter is considered large as compared with the radius. Blood flow is obtained, expressed in m^3/s and in ml/s; flow velocity in m/s and in mm/s; shear stress on the arterial wall and Reynolds number.

The option «physical parameters» in the menu shows a table containing the properties of the blood vessels.

Fig. 5. *Velocities simulator*

In the fork option from the menu, we have a flow Q1 in one vessel and a flow Q2 in each branch. Pushing the button, the flow in the principal branch and in each branch is calculated, as well as the angle of the fork.

Fig. 6. *Flow simulator in forks*

When choosing the basic cardiovascular Physiology button, several modules are accessed. One of them shows the elastance path corresponding to the left ventricle systole.

Fig. 7 corresponds to this module interface, where two register windows can be seen. The left frame shows a plot of elastance versus time. The right one shows the contraction velocities plot. Each process corresponds to the exact contraction times of the left ventricle. The amplitudes determine the average ventricular and aortic pressures of a healthy adult.
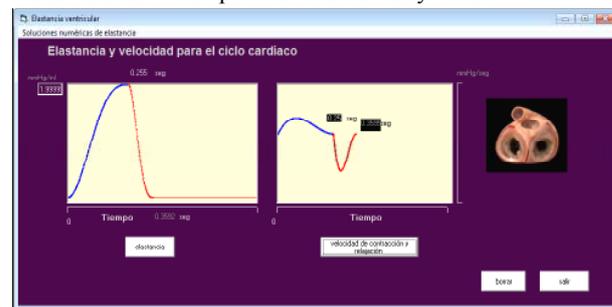

Fig. 7. *Elastance during a whole cycle*

This screen shows the elastance calculation, as well as the contraction velocities and cardiac relaxation. To the right a heart animation is shown. The lower buttons from each register plot the elastance and the velocities of the cardiac cycle. The numeric calculation of the left ventricle pressures, the proximal and distal aortic pressures, and the corresponding blood flow, are shown at the user interface in *Fig. 8*.

Three oscilloscope windows are shown. The greatest is in the lower central region It is recorded, when performing a simulation for the aortic and left ventricular pressures.. The oscilloscope window at the upper left shows the plot of the blood flow, whereas the upper right window shows the plot of velocities used to calculate the previous registers.
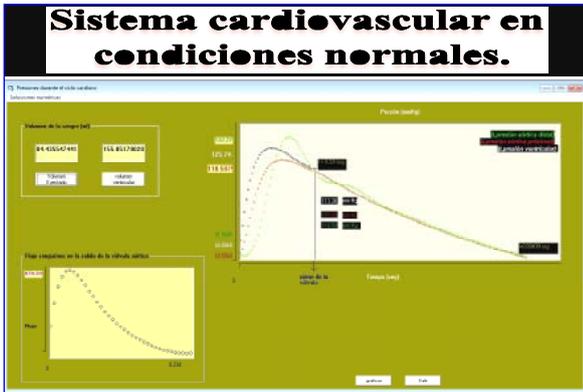
Fig. 8. *Interface showing traces of blood flows and pressures*

Now we can move to the simulation of some pathology. For example the one of aortic stenosis. It simulates the circuit shown in the following figure:
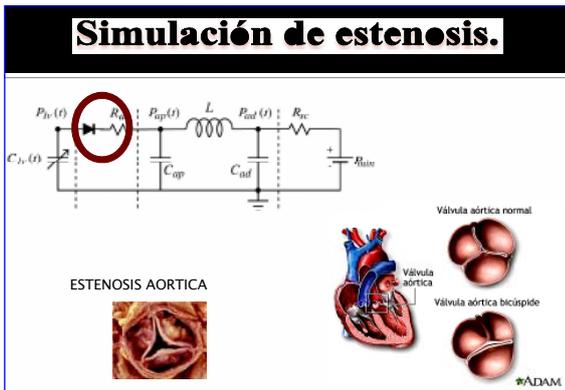

Fig. 9. *Modelling stenosis, now the aortic resistance is variable*

The value of aortic resistance is modified by the scrollbar, visualized at the left frame. The option "ventricular pressure vs proximal aortic pressure plots these functions for each stenosis level, and the screen shows the state of the aortic valve. The outline describes the difference between a healthy and a sick valve.
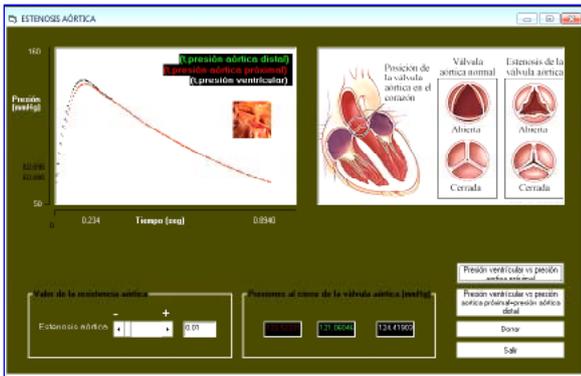

Fig. 10. *Simulator screen. Healthy valve*

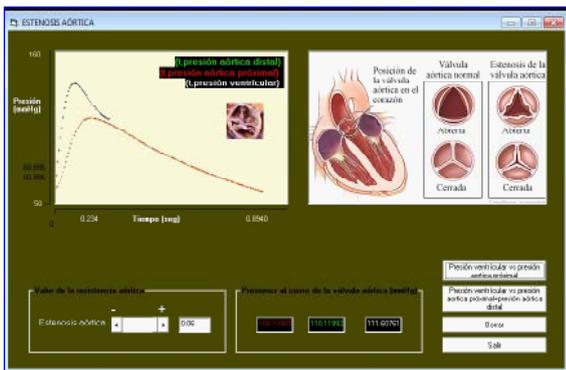Pressures at valve closure are shown in another frame.

Fig. 11. *Severe stenosis*

Finally, a first version of the simulator of hypertension is displayed.

Immediately at the start of the screen, it begins a video about high blood pressure. It shows the behavior of the distal aortic pressure for each value of resistance
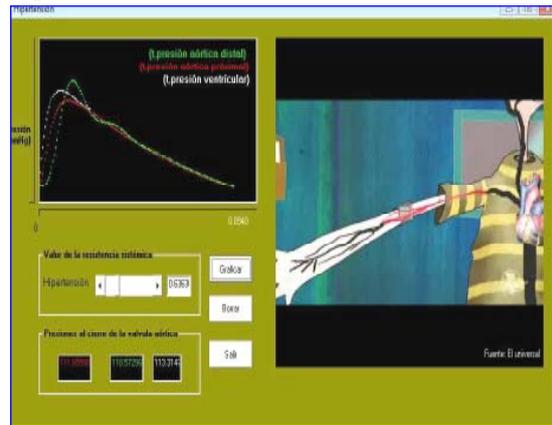

Fig. 12. *High blood pressure*

4. CONCLUSIONS

In this paper we present the results of the design and development of different cardiovascular simulators. These simulators allow the study of basic cardiovascular physiology and an introduction to hemodynamics. They are important didactic tools in medical education. A combined effort of specialists in different research areas including physics, medicine, mathematics, computer science, biophysics and education was necessary to design the simulators.

To simulate the behavior of pressure and blood flow, validated mathematical models previously published were used. The range of values for the variables is the ones reported in medical literature, and correspond to average values of adult healthy patients.

The calculated output values of the variables fall within this range; therefore the simulations do not belong to a particular patient.

The functioning of the simulator was validated when the calculated values of the variables were found within the reported value.

In this simulator some physiopathological aspects were included, i.e. stenosis of the aortic valve. All alterations allowed by the simulator correspond to qualitative conditions from lower to higher pathological intensity. They are not exact values of a particular clinical case. The user, preferentially a medical student, can observe the changes that occur after altering the input values of the variables. For the stenosis of the aortic valve, the result is an increase in the electric resistance representing the valve.

The simulators are designed for teaching, introducing the student to medical topics, where learning is improved by tools that facilitate reasoning.

The simulators developed here are useful in teaching and learning the cardiovascular system. Three different items were covered:

1) some basic aspects of hemodynamics;
2) some parts of the circulatory system were simulated;
3) general physiopathological aspects are treated.

Our future plan is to test the simulators in the medical curriculum and develop simulators for specific pathologies, where the student is involved in the treatment of a virtual patient. An impyrtant example, where the relationship between blood flow, arterial walls and bifurcations, venous return, etc. are considered is the case of the aneurism. This simulator is currently under developement.

References:

1. Westerhof N. Snapshots of hemodynamics / N. Westerhof, N. Stergiopulos, M.I.M. Noble. – New York : Springer, 2010.
2. Guarini M. Modelaciyn y circulaciyn cardiovascular / M. Guarini // Departamento de Ingenierнa Elйctrica, Pontificia Universidad Catylica de Chile, S/N. – 2010.
3. Reduced cardiovascular models / J. Peiro, A. Veneziani, L. Formaggia, A. Quarteroni, A. Veneziani (Eds) // Cardiovascular Mathematics. – Milano, Italia, 2009. – P. 347-394.

4. DrAlessandro-Martнnez A. Modelaje matemбtico y simulaciуn computacional del pulso ayrtico y de la eyecciуn ventricular / A. Dr Alessandro-Martнnez, R. Silva-Bustillo A. Sutil-Rosas // Congreso Latinoamericano de Bioingenieтна. – La Habana Cuba, 2001.

5. Westerhof N. The arterial Windkessel en Snapshots of Hemo-dynamics / N. Westerhof, N. Stergiopulos, M.I.M. Noble. – 2 ed. – New York : Springer, 2010. – P.173-181.

6. Fung Y.C. Biomechanics circulation / Y. C. Fung. – 2 ed. – New York : Springer, 1997.

7. Mazundar J. Biological fluid mechanics en An Introduction to mathematical physiology & Biology / J. Mazundar. – 2 ed. – New York : Cambridge University Press, 1999.

8. Biofluid mechanics en Medical Physics and Biomedical En-gineering / B.H. Brown, R.H. Smallwood, D.C. Barber, P.V. Lawford, D.R. Hose. – Great Britain : Taylor & Francis Group, 1999. – P. 46-75.

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ПРОЕКТУВАННЯ ТА РОЗРОБКА КОМП'ЮТЕРНОГО ТРЕНАЖЕРА БІОФІЗИЧНИХ ПРОЦЕСІВ У НАВЧАННІ МАЙБУТНІХ ФАХІВЦІВ

У статті описані методичні рекомендації для майбут-ніх фахівців щодо використання математичних моделей фізичних процесів. Це представлено через комп'ютерну програму-тренажер для формування професійних компе-тенцій майбутніх фахівців з провідної галузі.

Ключові слова: студенти, моделі фізичних процесів, засоби навчання, комп'ютерна програма-тренажер.

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ПРОЕКТИРОВАНИЕ И РАЗРАБОТКА КОМПЬЮТЕРНОГО ТРЕНАЖЕРА БИОФИЗИЧЕСКИХ ПРОЦЕССОВ В ОБУЧЕНИИ БУДУЩИХ СПЕЦИАЛИСТОВ

Обучение студентов может быть значительно воспри-имчевее, если поддерживается с помощью компьютерных программ-тренажеров. Программы-тренажеры позволяют студентам изучить каждую переменную в конкретных усло-виях поставленной задачи, в исследованиях по их измене-ниям. Эти инновации имеют жизненно важное значение для изучения сердечно-сосудистой системы у студентов-медиков, а также для студентов смежных специальностей.

Ключевые слова: компьютерная программа-тренажер, студент, средства обучения.