Результати експериментальних досліджень демонструють позитивну динаміку нормалізації біомеханічних параметрів, що підтверджує доцільність застосування інструментального підходу у процесі клінічного ортезування. Такий підхід сприяє скороченню термінів реабілітації, покращенню функціонального стану пацієнтів і зменшенню ризику повторних ушкоджень.

Ключові слова: ортезування, навантаження, стопа, коефіцієнт розвантаження, цикл кроку.

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OPTIMIZATION OF THE CONTROL SYSTEM OF AN AUTOMATED TRANSFEMORAL PROSTHESIS FOR PEOPLE WITH GAIT DEFECTS

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Introduction

Modern realities make the problem of restoring the functionality of the musculoskeletal system for people who have lost it due to diseases, injuries and limb amputations very relevant [1].

One of the most important parts of the musculoskeletal system are the legs. The working lower limbs play an important role in human health and ensure their self activity. Sufficient motor activity of the legs improves the functioning of the cardiovascular system and maintains the tone of the entire musculoskeletal system. Therefore, in prosthetics, restoring their functionality is one of the priority tasks. There are five types of lower limb amputation, and among them, transfemoral amputation is one of the most common and severe.

In such cases, one of the main goals is to restore the functionality of the knee joint. In this regard, digital knee joints with complex control systems that adapt to different situations are becoming increasingly widespread. Their main disadvantage is their low availability. Also, the control systems of these prostheses still do not cover a large part of the cases when people have certain temporary or permanent gait defects.

Therefore, the purpose of this work is to optimize the transfemoral prosthesis control system taking into account gait defects of people with lower limb movement disorders.

Material bases and elements of the control system

First, in order to optimize the control system, it is necessary to choose a prosthesis on which direct optimization will take place.

1. Frame and control mechanism

The design proposed in [2] can be taken as a basis for further improvements. This design is an automated digital uniaxial knee prosthesis.

Transfemoral prostheses, in addition to being divided into digital and mechanical, can also be divided into single-axis and multi-axis. The advantage of multi-axis prostheses is increased stability when used even without additional stabilization elements. They are also much more comfortable to walk in comparison with mechanical single-axis ones. However, the use of a single-axis prosthesis will significantly reduce its weight, freeing up space for the control and stabilization system, and allowing more comfortable use for people with weakened muscles.

It was decided to replace the proposed design of the variable liquid damper in [2] with the magnetic damper Bohai MR C39-40B01. The choice was due to its lower weight, greater reliability and ease of use. This shock absorber has a total length of 240 mm and a cylinder diameter of 39 mm [3]. It is able to withstand a load of up to 1350 N and is regulated by a current of 0-1.5 A [3]. The operating voltage is +24 V.

2. Sensor system and control system

In work [2], an element base for the control system and sensor system is proposed, consisting of: an FX29 strain gauge, a GY-521 module, an Atmega328 microcontroller as part of the Arduino UNO development board, an encoder, a Tower Pro MG90S servo drive, a DC Mini560 power adapter, and NCR18650 batteries.

It was determined that most of them would not be used, and therefore the decision was made to replace them with options with better parameters.

Two GY-521 modules will be used as the main elements of the sensor system. This module is built on the basis of a three-axis gyroscope MPU6050, which also has a three-axis accelerometer built in. Also, this module has a temperature sensor, the functionality of which is irrelevant for this control system.

Despite this, the main advantage of using the GY-521 module is the ease of its implementation in a control system built on the Arduino platform. We work with them via I2C, using addresses 0x68 and 0x69. In addition, this module has a built-in 5 V converter, which allows direct power supply from the development board. Also, they are small enough (dimensions 20 x 16 x 2 mm) that they can be easily installed on any necessary surface of the prosthesis. Their main function will be to set the angles of inclination of the prosthesis parts for further determination of the stages of movement.

The next sensor chosen is the FSR-402, as a resistor, whose resistance changes with the load. A simple and affordable alternative to the FX29 strain gauge. It can be connected directly to Arduino boards. This sensor is installed in such a way as to fix the contact of the leg with the ground. As batteries for the prosthesis, you can choose the proposed NCR18650. 7 batteries connected in parallel will provide the necessary voltage for the damper to operate.

The main control element is a board based on the Atmega328 processor. The use of the Arduino UNO development board is not necessary, since for all the tasks set, the Arduino Nano board based on the same processor will be sufficient. This board is a much more compact and cheaper solution. This board has a clock frequency of 16 MHz, 2 KB of SRAM, 32 MB of flash memory and 1 KB of EEPROM. In addition to simple analog and digital ports, it has I2C ports, 3.3 V power supply and two ports for communication via Bluetooth modules. A significant difference is that this board processes floating points programmatically. Arduino UNO will control a variable magnetic damper via an IRF540N transistor with a threshold voltage of 4 V and a maximum allowable input-output voltage of 100 V. Also, for the entire sensor system and processor to work, it is necessary to connect Arduino UNO to 24 V via the Mini MP1584 DC-DC 3A module. This converter is suitable because it is able to lower the voltage from 4.5 V - 28 V to 0.8 V - 20 V in direct current. The method of connecting the selected control elements is shown in Fig. 1.

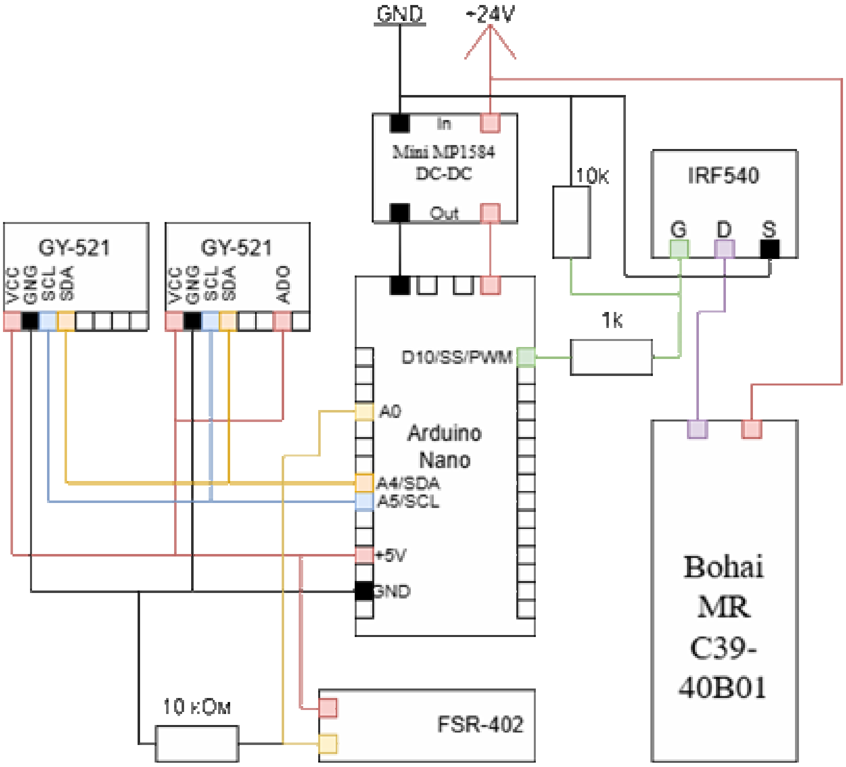

Fig. 1. Connection diagram of the control system and the sensor system

Control system software

To create a correctly working control system, it is necessary to understand the parameters, which will be managed. The sensor system discussed above has as its main task the tracking of angles of inclination of parts of the prosthesis during walking.

Therefore, based on the compiled connection diagram of the control system, a prototype of the sensor system was created.

Prototype development

A part of the system consisting of Arduino Nano and two GY-521 modules was selected. Their connection scheme remains unchanged. A battery (+9 V) was used as a power source. For convenient data transfer to a personal computer, a Bluetooth module Hc-05 was used. Arduino IDE, Python and Microsoft Excel were used to write programs and process data. When installing the second gyroscope, +5 V was supplied to its AD0 port to change the address of this sensor. Thus, simultaneous uninterrupted use of both gyroscopes through one I2C port is ensured.

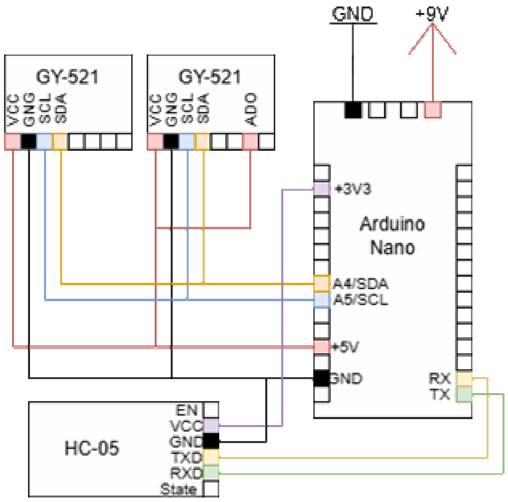

Fig. 2. Connection diagram of the prototype sensor system

The angle of inclination of the module is calculated by integrating the rotation speed according to the equation:

$$A = \int_0^{k \cdot T_s} R \cdot dt\,,$$

where A – angle of inclination; R – rotation speed; k – iteration number; Ts – iteration time.

Also, the angle of inclination can be found using the accelerometer built into the module according to the equation:

$$\theta x = \arctan\left(\frac{Acc_x}{\sqrt{Acc_y{}^2 + Acc_z{}^2}}\right),$$

where Acc – axial acceleration.

Both methods have certain flaws. Finding angles using accelerometers makes the system vulnerable to noise and small movements. In turn, finding using a gyroscope leads to the accumulation of errors with each iteration. To eliminate these inaccuracies, a Kalman filter was used, which allowed combining these two methods.

The first result was the re-invention of the equation for the integration of fluidity in rotation, which appeared as follows:

$$A_{kalman}(k) = A_{kalman}(k-1) + T_s \cdot Rate(k)\,.$$

We assume that the resulting angle is not the final value, but only a prediction. In this case, this prediction will have an uncertainty, which can be described as follows:

$$U_A(k) = U_A(k-1) + T_s^2 \cdot 4^2\,,$$

where U – uncertainty.

So, now we can calculate a new prediction for the angle:

$$A_{kalman}(k) = A_{kalman}(k) + G_{kalman} \cdot (A(k) - A_{kalman}),$$

where $A(k)$ – angle measured using accelerometers.

In turn

$$G_{kalman} = \frac{U_A(k)}{U_A(k) + 3^2}.$$

The last equation is necessary to update the uncertainty for the next prediction. This way, a signal is obtained that is close to the real one and free from extraneous noise.

The HC-05 module was configured to a data transfer rate of 115200 baud as follows. To prevent errors when transferring data from the Arduino IDE to Python, start and end characters were set. The data is transferred to a suitable python program, which accepts it, discards the accompanying characters, generates detailed real-time graphs and records the data in a Microsoft Excel file. From there, they can be freely taken for evaluation and analysis.

The prototype consists of two parts: the upper and lower sensors. Along with the upper sensor, there is also an Arduino UNO board, a Bluetooth module HC-05 and a power supply. All elements are securely fixed to minimize errors and breakage. They are fixed to the leg using adhesive tapes. The sensors are located above the knee and ankle joints, deployed as shown in Fig. 3, with the sensor axis X set parallel to the knee flexion axis.

Fig. 3. Photo of the prototype mounted on the leg .

Study of angular parameters of the leg during healthy and defective gait

To form a correct idea of the obtained data, it is necessary to understand what the gait cycle consists of and what the trajectory of the leg is during healthy gait. The classification of stages is shown in Fig. 4. It is also useful to focus on the existing information on the dynamics of movement of parts of the leg during healthy gait. Similar data are given in [5, Fig. 7, Fig. 8], which illustrates the features of the dynamics of changes in the angle of inclination of the thigh relative to the pelvis and the lower leg relative to the thigh. With respect to them, the quality of the sensor system can be established.

In turn, Fig. 5 of our work shows new data measured using a prototype sensor system. If we compare the dynamics of the change in the angle of inclination of the thigh relative to the ground, it is in the range of maximum and minimum deviations according to Fig. 5. Despite this, it will not be possible to make a similar comparison for the graph of the movement of the lower leg, since the new data are also measured relative to the ground, unlike those given in the study [5, Fig. 7, Fig. 8]. Among other features of the new measurements, we can note a small shock that occurs when the foot contacts the ground. Now, this sensor system can be used to obtain the dynamics of the movement of a defective gait and analyze it for further input into control systems.

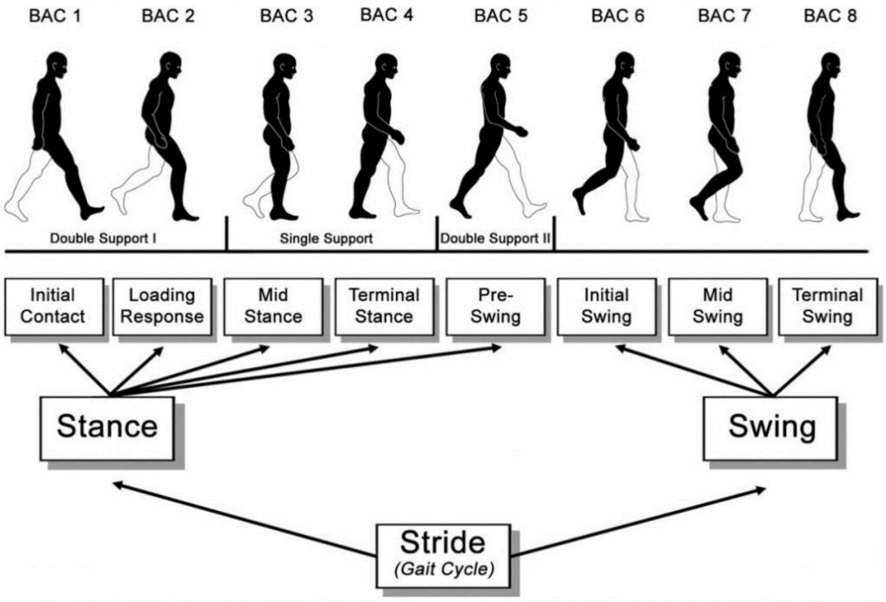

Fig. 4. Stages and phases of human gait [4].

The most common indication for amputation in our time is serious injury. Typically, the causes of such injuries can cause a number of other health problems, ranging from direct damage to the musculoskeletal system to severe neurological injuries.

In view of this, the following gait defects will be examined: unsteady gait; antalgic gait; Trendelenburg gait [6] and Parkinsonian gait [7].

Unsteady gait occurs due to significant weakening or complete paralysis of the gluteus maximus muscle. As a result, the person has significantly impaired control over the thigh, which is why he has to compensate for the inability to raise the limb by tilting the body. In other words, the affected person swings the leg with his body in order to open the knee and take a step.

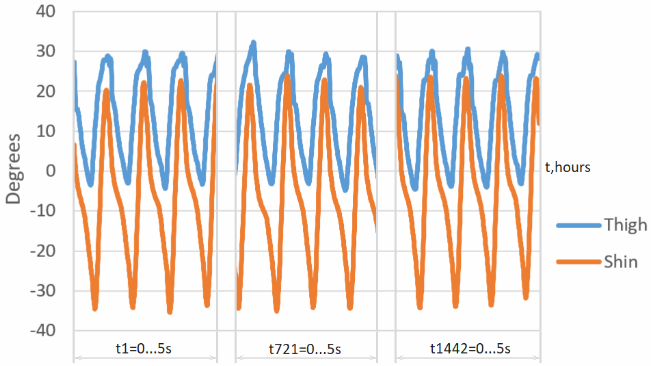

Fig. 5. Dynamics of changes in the angles of inclination of the lower leg and thigh relative to the ground.

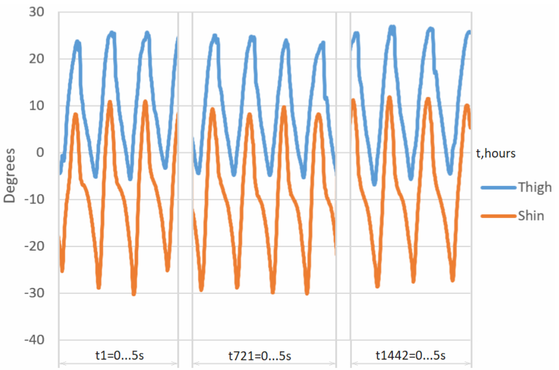

Fig. 6. Dynamics of changes in the angles of inclination of the lower leg and thigh in people with a wobbly gait.

This defect is similar to the problem with too heavy prostheses mentioned above. While in the case of weakness the muscle can still be rehabilitated, in the case of paralysis this defect is most often permanent. The study of the dynamics of leg movement is shown in Fig. 6. The graph shows a decrease in the amplitude of movements by 20 degrees for both parts of the body. What also distinguishes this defect is the greater uniformity of hip movements, which is expressed in the increase in the symmetry of the graphs of the 1st cycle relative to their tops.

Antalgic gait occurs due to severe pain in the affected leg. They can be caused by discomfort due to an incorrectly made socket, or due to a recent injury. Such a gait is characterized by a shorter support phase of the gait cycle for the affected limb. In some cases, the affected limb is constantly half-bent, which is why the person has to lean the torso more to touch the ground. This problem is more often treatable and the control algorithm for it will be used temporarily. The dynamics of the movement of the leg with antalgic gait are shown in Fig. 7.

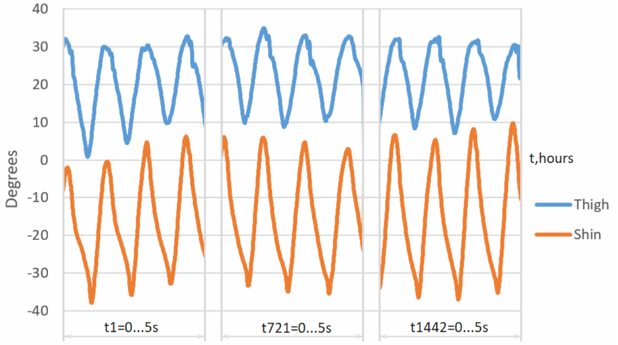

Fig. 7. Dynamics of changes in the angles of inclination of the shin and thigh in people with antalgic gait.

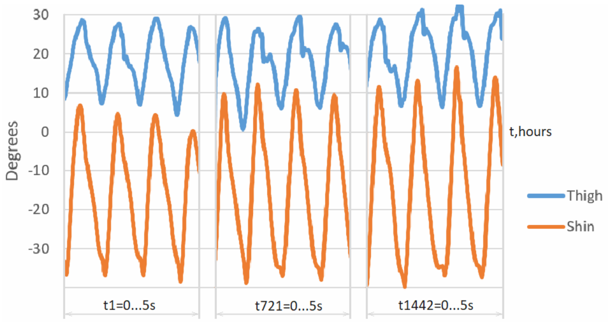

Fig. 8. Dynamics of changes in the angles of inclination of the lower leg and thigh in people with Trendelenburg gait

Because a person moves with one leg bent, the range of motion becomes even smaller. Also, when adjusting prostheses, one can rely on the fact that the range of motion of the lower leg and foot almost does not overlap.

Trendelenburg gait occurs due to significant weakening or complete paralysis of the gluteus medius muscle. As a result, in the middle of the support phase of the affected leg, the unaffected leg bends, and the pelvis on that side sinks down slightly. This is due to the fact that the gluteus medius muscle on the affected side is unable to maintain the vertical position of the body. As in the case of a wobbly gait, this defect is treatable when weakened, less often when paralyzed. The dynamics of leg movement with Trendelenburg gait are shown in Fig. 8. Trendelenburg gait can be identified by looking at the significant difference between the amplitudes of the hip and shin oscillations. Another secondary sign is an unstable hip stance.

Parkinsonian gait occurs as a result of the development of Parkinson's disease, which is characteristic of older people, genetically susceptible people and people with a frequently injured central nervous system. This disease affects both legs, causing short steps on almost always bent knees with a significant forward tilt of the body. The main role is played by the movement of the hips, since the knees are almost immobile during this gait.

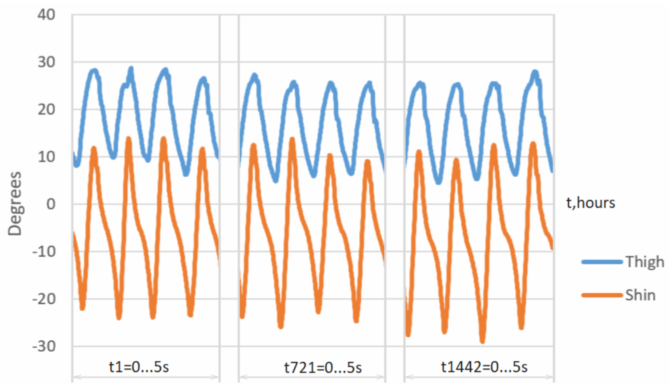

Fig. 9. Dynamics of changes in the angles of inclination of the lower leg and thigh in people with Parkinson's gait.

This disease is incurable. At the moment, doctors are only able to slightly alleviate its symptoms. The dynamics of leg movement with Parkinson's gait are shown in Fig. 9. The graph clearly shows a larger number of small steps with much smaller movement amplitudes, which almost do not overlap each other.

Conclusions

So, taking into account the selected data, such as elements: Arduino Uno; 2 GY-521; HC-05 module; 9 V battery; power consumption 4.5 W; weight 185 g, as a result of the optimization performed using heuristic methods, the minimization of the impact of noise on the system and the maximum approximation of the output signal values to the real values of the leg inclination at a speed of 100 measurements per second were obtained.

The important topical issue of improving control systems for digital transfemoral prostheses was considered in this work. Using the example of an existing transfemoral prosthesis design, its improvement was carried out in both the mechanical module and the electronic module. Based on the proposed improvements, a sensor system was created, calibrated and tested, which is capable of detecting gait abnormalities. Using the sensor system, various types of gait defects were investigated and clear differences between them were established.

Key words: transfemoral prosthesis; control system’s optimization; Parkinson's disease; antalgic gait; dynamics of leg movement; sensor system.

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КОМП’ЮТЕРНО - ІНТЕГРОВАНА СИСТЕМА ЛАЗЕРНОЇ МЕДИЦИНИ

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Сучасна лазерна медицина охоплює широкий спектр терапевтичних , хірургічних і реабілітаційно-відновлювальних процедур, в яких ключову роль відіграє точність дозованих параметрів, контроль і стабільність параметрів лазерного випромінювання [1]. Створення комп’ютерно-інтегрованих систем для автоматизованого управління такими процесами дозволяє не лише покращити результати лікування, а й забезпечити безпеку та комфорт пацієнтів [2].

Використання комп’ютерного аналізу біосигналів, таких як фотоплетизмограма, дозволяє оцінити адаптаційний стан пацієнта в реальному часі [3]. Це дає змогу лікарю-фізіотерапевту індивідуально налаштовувати параметри лазерного впливу, що підвищує ефективність терапії та знижує ризик ускладнень. Оптимізація параметрів лікування, завдяки комп’ютерному моделюванню можна підбирати оптимальні режими лазерного впливу ще до початку процедури [4]. Це особливо важливо при магнітолазерній терапії, де точність дозування світлового випромінювання критично важлива для досягнення бажаного терапевтичного ефекту [5]. Підвищити безпеки та контроль параметрів впливу дозволяє інтеграція комп’ютерних систем, за рахунок здійснення постійного моніторингу стану пацієнта під час фізіопроцедури, що забезпечує своєчасне виявлення та корекцію можливих відхилень [6]. Це сприяє підвищенню захищеності і безпеки пацієнта та якості лікування.

Зростає популярність процедур, які не потребують глибокого хірургічного втручання. Це дозволяє пацієнтам швидше відновлюватися та зменшує ризик ускладнень [7]. Наприклад, лазерне шліфування шкіри обличчя забезпечує омолодження і відновлення поверхневих шарів шкіри без тривалого періоду реабілітації. Розширення застосування лазерних технологій. Лазери використовуються не лише в хірургії, терапії, косметології, але й у стоматології, гінекології та інших напрямах медицини [8]. Це сприяє більш точному та ефективному лікуванню різноманітних захворювань. Інтеграція з комп’ютерними системами керування в лазерних апаратах, що оснащуються