Robotic control system for lower limb rehabilitation with force feedback

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Abstract. The paper proposes the structure of the robotic system (RS) control system with and without the patient’s force feedback, describes the conditions for using each of them. The method for improving data reliability on applied forces using strain gauge readings in the absence of patient activity was developed. The method of motivating a patient to perform “correct” movements thus “assisting” the mechanism, which consists in changing the speed of movement to create an impression that the mechanism moves due to patient’s efforts was developed. Mathematical RS model was constructed, graphs reflecting the accuracy of the specified trajectory were obtained.

1. Introduction
Rehabilitation of patients suffering from musculoskeletal diseases of varying severity is one of the most difficult tasks of modern medicine, the relevance of which is undeniable. Despite a careful attitude of people to their health, the number of diseases associated with impaired cerebral circulation that lead to a stroke has not decreased in recent decades [1]. The development of such diseases in some cases results in complete or partial immobilization of a patient thus requiring special care from third parties. At the same time, the restoration of limbs normal functions can be significantly accelerated through appropriate rehabilitation measures. One of such efficient measures leading to significant improvements in the condition of patients with impaired musculoskeletal functions is the use of robotic physiotherapy [2]. Various orthopaedic rehabilitation systems and devices are proposed to treat patients with musculoskeletal diseases [3-5]. Among other studies, the work [6] proposes a system for the RS-based rehabilitation of lower limbs (Fig. 1) using active 3-PRRR mechanism [7] and a passive orthosis. The active 3-PRRR parallel mechanism provides the required rotation angles of the patient’s joints for rehabilitation, while the passive orthosis is used to support the patient’s lower limb.
The used 3-PRRR mechanism (Fig. 2) has three independent kinematic chains $A_iB_iC_iD_i$, each of which has one prismatic joint equipped with the electric drive, and three rotary joints. The connection of joints in a chain and chains with a movable platform provides the latter one with three translational degrees of freedom.

The purpose of the study was to develop the RS control structure and to ensure its modeling in order to check the operability of the system. The efficiency of the concept of encouraging the patient’s efforts was also experimentally verified.

2. Development of a structural control scheme for the robotic system
The proposed complex is controlled by organizing the necessary movement of its active component. In order to develop the control system, it is necessary to establish a connection between the input and output coordinates of the 3-PRRR mechanism. Fig. 2 shows that the coordinates $x_p, y_p$ and $z_p$ of the
P point are determined respectively by the values \( r_1, r_2 \) and \( r_3 \), i.e. the coordinate \( x \) of the \( A_1 \) point, the coordinate \( y \) of the \( A_2 \) point and the coordinate \( z \) of the \( A_3 \) point: 
\[
x_p = r_1 - dx = x - dx, \\
y_p = -(r_2 - dy) = y + dy,
\]
where \( dx \), \( dy \) and \( dz \) – constants that determine the position of the P point relative to the center of the movable platform. Since each output coordinate is uniquely determined by the corresponding input and the change in the state of prismatic joints results in the same change in the position of the P point, in order to reach, for example, the given coordinate of the P point along the \( x \) axis, it is enough to reach the corresponding coordinate of the \( A_1 \) point along the same axis. As a result, the task of controlling the output coordinates is reduced to the task of controlling the inputs [8], and the feedback on the position of prismatic joints are, in fact, the feedback on the position of the P point (Fig. 3, dashed line).

The scheme in Fig. 3 includes three independent contours that regulate the state of prismatic joints (positions of points \( A_1, A_2, A_3 \)). For control it is necessary to calculate the setpoints for these contours based on the specified trajectory of the P point. The use of the above scheme allows controlling the RS in order to ensure movement along the specified trajectory at a specified speed. It should be noted that such control is quite sufficient for the rehabilitation of a patient with a deep impairment of motor functions when he is not able to control the movement of his limbs. During rehabilitation, the ability to move gradually returns to patients.

The basic idea is as follows: if a patient makes or tries to make “correct” movements thus “helping” the mechanism, the movement is accelerated so that the patient gets the impression that the mechanism is moving due to his efforts. To determine the patient’s forces, a system of strain gauges can be used to define the projections of force on different coordinate axes. First, it is necessary to record the initial trend of their readings when performing the same movements, but when the patient does not perform any activity. This trend can then be used as a setting for the contour (Fig. 4) to adjust the patient’s forces, i.e. the scalar projection of the force vector measured by the sensors to the platform speed vector. The use of such a projection makes it possible to take into account the component of the patient’s force, which coincides with the current direction of movement (or is opposite to it). If the control system detects positive patient activity compared to stored data, the regulator will increase the frequency of task change (and the retrieval of retrospective tensometer readings). Changing the speed of movement will lead to changing the readings of sensors (making “correct” movements will become more difficult). As a result, the speed corresponding to the actions of the patient will be selected.

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**Fig. 3.** Diagram of RS control system without patient’s effort feedback.
3. Simulation of the control system
The model developed in the MATLAB environment (Fig. 5) consists of three subsystems: SUB-1, SUB-2 and SUB-3 corresponding to three parallel chains connected by one end to a fixed base formed by three mutually constant-coular rods, and by the other – to the Movable platform. The tasks on the state of drive links are transmitted to the corresponding subsystems. Each of the above subsystems (Fig. 6) comprises a kinematic circuit including a Prismatic Joint and three Revolute Joints. The black color on the figure shows the control system, lilac – physical signals, blue – electrical part, light green – rotating elements, dark green – prismatic elements, gray – kinematic circuit.

Fig. 4. Diagram of RS control system with patient’s effort feedback.

Fig. 5. Mathematical model diagram.
Information on the state of the prismatic joint is used as a feedback signal in its position control contour. The control action generated by the regulator is divided into two components: absolute value, which in the Controlled PWM Voltage unit is converted into a width-modulated pulse electric signal, and the action sign, which is used in the H-Bridge unit to change the polarity of the output voltage and the reverse of the DC Motor connected to it. The rotational motion of the motor shaft is converted into a translational motion that changes the position of the prismatic joint through a Leadscrew transmission. Linear speed and applied force are matched using Ideal Translational Velocity Source and Ideal Force Sensor units.

The passive orthosis (Fig. 5) is attached to the P point through a ball joint. The orthosis is a sequential RRRS mechanism and makes the patient’s lower limb fixed on it rotate in the joints: by the knee – Articulatio Genus and by the thigh – Articulatio Coxae1, Articulatio Coxaec2 (rotation in two directions).

The resulting model allows testing the possibility of the RS motion along the trajectories necessary for the rehabilitation of lower limbs. One of the rehabilitation programs is the development of leg movements during walking. The analysis of literature sources on this topic showed some methods for reproducing a person’s gait depending on his anthropometric characteristics, in particular the size of the thigh and the ankle. The results of such simulations are presented, for example, in [9]. The functions of angles to the vertical of the thigh (α) and the ankle (β) of the supporting and moving legs respectively can be approximated by the ratios:

\[
\alpha_1 = \beta_1 = 0.198 - 0.945t + 0.954t^2 - 1.508t^3 + 0.710t^4 - 0.568t^5
\]
\[
\alpha_2 = -0.198 + 4.546t - 17.540t^2 + 45.266t^3 - 53.113t^4 + 5.413t^5
\]
\[
\beta_2 = -0.198 - 3.036t + 18.534t^2 - 38.374t^3 + 30.526t^4 + 5.413t^5
\]

where \( t \) – time from the moment of the supporting leg change (step duration – 1s).
To describe the leg abduction in the thigh joint, an $\gamma$ angle was also introduced: $\gamma_1 = 0, \gamma_2 = \sin(2\pi t) \cdot \pi / 36$. The trajectory of the P point is determined by the ratios:

$$x_p = x_0 - (l_c + l_f) \sin \gamma, \quad y_p = y_0 + (l_f \cos \alpha + l_c \cos \beta) \cos \gamma \cos \beta, \quad z_p = z_0 + l_f \sin \alpha + l_c \sin \beta,$$

where $x_0, y_0, z_0$ - coordinates of the thigh joint, $l_f$ and $l_c$ - length of the thigh and ankle, respectively. As a result of the simulation, the trajectory of the P point was obtained (Fig. 7), the maximum dynamic error was less than 3 mm (Fig. 8). The task (blue) and result (orange) graphs are almost the same.

![Fig. 7. P point motion path in coordinates: a) yz, b) xz.](image1)

![Fig. 8. Dynamic error by coordinates: a) x, b) y, c) z.](image2)

4. Simulation of the control system responsive to patient effort dynamics

To implement the control system in MATLAB environment according to Fig. 4, the diagram shown in Fig. 5 was modified to add the possibility of changing the task change rate for the prismatic joint position control contours (Fig. 6) depending on the patient’s forces. For this purpose signals from outputs 1, 2 and 3 of the subsystem are supplied to the inputs of subsystems SUB-1, SUB-2 and SUB-3, which change the task change rate for position control circuits depending on patient’s forces. The 1-D Lookup Table units contain the same tasks as in the diagram in Fig. 5, but make them be replaced not in time, but depending on the input signal, which increases at a rate determined by the signal value at the lower input of the adder Add1. If this signal is negative or zero, the minimum task change rate is set. A positive signal increases the growth rate of the signal at the output of the Integrator unit. The maximum acceleration of the task change is defined by a time constant of the lower frequencies filter Transfer Fcn1.
Random Number, Sine Wave, Transport Delay and Add units are used to simulate the patient’s forces at each of the coordinates (difference in the readings of the strain gauges during activity and rest). Low Pass Fcn filters are used to smooth the sensor readings. The rate at which the tasks are changed for each position control contour is determined by the Derivative units. The Product, Square, Add1, Add2, Sqrt, and Divide units are used to calculate the scalar projection of the force vector measured by the sensors by the platform speed vector: \( F_v = F \cdot v / \| v \| \). Two patient behaviors were modeled during the experiment (Fig. 9). The figure shows the normalized scalar projection graphs of the patient force vector by the platform speed vector (orange) and the relative task change rate graph (blue). The first 100s the patient is inactive, and as a result, his limb moves with the mechanism at a minimum speed. The following 100s the patient is active and the direction of his efforts generally coincides with the direction of movement of the movable platform.

![Fig. 9. Normalized force graphs](image)

The trajectory of the P point obtained as a result of the simulation is identical to that shown in Fig. 8, the dynamic error when moving at a lower speed is less (Fig. 8 and Fig. 10). The maximum dynamic error still does not exceed 3 mm.

![Fig. 10. Dynamic error at the movement from a variable speed by coordinates: a) x, b) y, c) z.](image)

5. Conclusion
The diagrams of the RS control system with and without patient’s effort feedback make it possible to control the mechanism during rehabilitation measures for patients with various degrees of motor dysfunction. The designed RS model allowed obtaining graphs reflecting the accuracy of a given
trajectory. The maximum dynamic error was 3 mm. It can be seen from the obtained graphs that the increase of the speed characteristics of the used drives by 2 times will reduce it to 1 mm. The method of motivating the patient to perform “correct” movements, which ensures his encouragement through the accelerating mechanism, was successfully tested on the model.

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