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MODELING A UNIVERSAL ACTIVE MODULE OF HUMAN MOTOR FUNCTION REHABILITATION DEVICE

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Structural and dynamic modeling of the universal active module of rehabilitation device with polymer-metal composite actuators for restoring the lost functions of the musculoskeletal system because of diseases of human motor neurons is performed. An investigation of an actuator's mechanical properties has also been done. A new approach for dynamic modeling of rehabilitation devices taking into account their joints' and links' flexibility and the effect of active natural skeletal muscles is proposed.

***Keywords*:** rehabilitation device, polymer-metal actuator, structural and dynamic modeling, flexibility of joints and links.

Introduction. While infantile paralysis in the high industrialized countries almost never occurs, in different parts of the world, people are still suffering from this disease. And other diseases of the central nervous system, pathological changes of the peripheral nerves caused by injuries, tumors, poisonings due to metabolic disturbances leave behind a group of patients with sequels of paralysis of the motor nerves. Paralysis leads to deformations of the growing child's skeleton, which complicates the further treatment [1]. Therefore, the use of orthosis is a valuable aid in the treatment of many of these patients. The design of new materials, particularly EAP-s (electroactive polymers) gives an opportunity to develop considerably new rehabilitation devices. The low weight, high flexibility, mobility, concentration of axes of links and joints are important for them. The aim of this article is to design a new type of orthosis which will provide all these characteristics.

An actuator and the actuator system design. For designing a rehabilitation device are selected PPY-metal coil composite actuators for which it is known that one fiber of the actuator with a diameter of 0.25 *mm* develops a force of 0.2 *N*, a bunch with 2500 fibers and 40 *mm* of diameter develops a force of up to 500 *N* (Fig. 1, 2). The control of these actuators can be implemented with varying of electric voltage, and the voltage-strain, voltage-force formulae are known.

PPY-metal coil composites may be used to replace the electro-mechanical actuators in a vast variety of application areas ranging from biomedical devices to soft actuators and sensor systems in modern humanoid robotics. It is one of the most promising active (smart) materials for developing new soft biomimetic actuators and sensors. The advantages of the PPY-metal coil composites include low driving voltage (<5 *V*), relatively high strain, soft and flexible structure.

These actuator bunches with their characteristics (40 *mm* of diameter, 500 *N* of force) fully comply with the requirements of rehabilitation devices. However, in case of motor neuron diseases, not all muscles included in the joints' movements and have some impact on the movement can be weakened (or hyperactivated). For balancing of these effects and for providing natural anatomical movements of joints, it is more convenient to create a system with a large number of actuators. By varying the voltage applied on the fibers, it is possible to increase or decrease the developing forces, and thereby provide the equal effects of extensional and flexional effects of muscles.

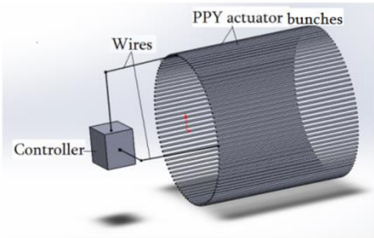

Fig. 1. Actuator system

Fig. 2. Fiber of actuator

Since actuators are working under the influence of external load, it is important that the forces developed by the actuators be greater in order to avoid undesirable strains. This condition is expressed by the comparison of the critical force determined by Euler and the force developed by the actuator under the influence of electrical current:

$$F_k \geq F_{cr},$$
$$\frac{\pi^2 EI}{(KL)^2} \leq \left(\frac{(1-2\nu)q^2}{4cV_k^2} + \sigma_k\right) \cdot S_k,$$
$$\sigma_k \geq \frac{\pi^2 EI}{(KL)^2 S_k} - \frac{(1-2\nu)q^2}{4cV_k^2}$$

by contributing D and H parameters and the maximum strain rate of actuators $(10\ \%/t)$ we derive:

$$\sigma_k \geq D - H(U \cdot t)^2,$$

where σ_k is the mechanical stress, U – the control voltage, t – the time of contraction of each actuator (2 *s*), $k = 1, \dots, n$ – the number of links and actuators of the exoskeleton, the *D* and *H* parameters are constant for a definite type and for given geometrical sizes of the actuator. As we can see from the Fig. 3, σ_k must be greater than $1.3 \cdot 10^7$ *Pa*.

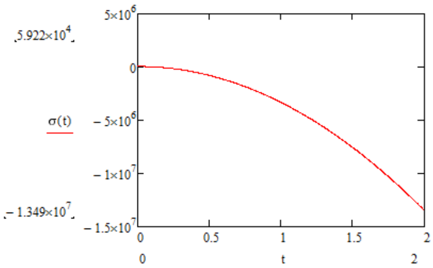

Fig. 3. Mechanical stress of each actuator (Pa)

Dynamic modeling. Let's consider the motion of the biomechanical system. The structure of which is given in Fig. 3 and 4. It consists of two parts which are connected to the body parts correspondingly creating a joint, an actuator system and a flexible joint.

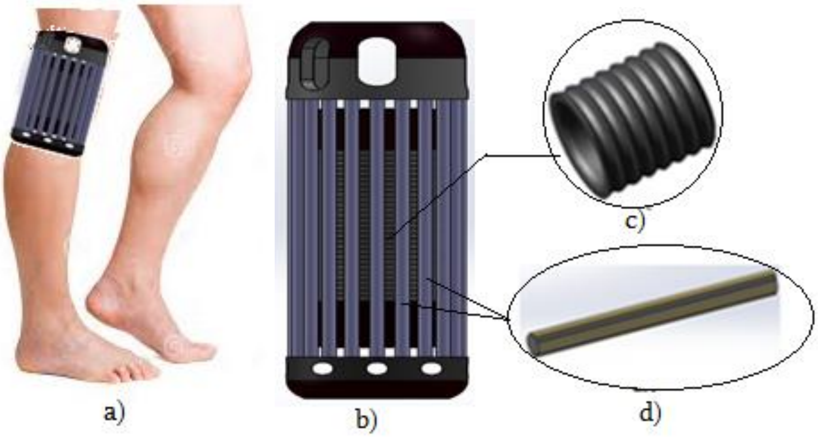

Fig. 4. a) Applicatin of orthosis, b) orthosis, c) flexible joint, d) actuator

Electroactive actuators are connected similar to natural muscles and are capable of changing their size and shape when an electrical signal is transmitted. The deformation of actuators is similar to natural muscles' deformation, except for one difference – they can work in both reduction and expansion modes. The activation and control of actuators are carried out through the change of electrical voltage values. The control can also be performed with enhanced brain signals. The flexible joint is located in the area of articular segment and provides stability, and due to accumulation of the elastic deformation energy, supports the system movements. The flexible joint also allows small translational movements of joints [2].

The dynamics is represented by the equations of motion of which the *i*-th equation can be written with the Lagrange formulation as follows:

$$\frac{d}{dt}\left(\frac{\partial T}{\partial \dot{q}_i}\right) - \frac{\partial T}{\partial q_i} + \frac{\partial P}{\partial q_i} = Q_i (i = 1)$$

with T and P respectively the total kinetic and gravitational energy of the mechanism, Q_i are the generalized forces associated with the generalized coordinates q_i. The dynamic modeling of mechanisms considering the links' and joints' elasticity were represented in our previous article [3]. As mentioned above, in the cases of motor neuron diseases, not all muscles are weakened, therefore they have some impact on the movements of the biomechanical system. For example, in the case of a small activity of the following muscles: semimembranosus muscle, rectus femoris, sartorius and etc, it can be calculated in the equations of system's motion as follows:

$$I_a(q)\ddot{q} + F_m(\dot{q}) + c(q, \dot{q}) + g(q) = \tau.$$

The muscles of the skeleton mainly have a form of a feather, so F_m will be determined with the following formula:

$$F_m = \frac{2bl \sin\alpha \cdot F \cos\alpha}{A},$$

where F is the developed force by one fiber, b – the thickness, l – the length, α – the feather angle, A - the cross section area of the fiber [4].

The simulation is performed with the program ADAMS. The models and results are depicted in the Fig. 5-7.

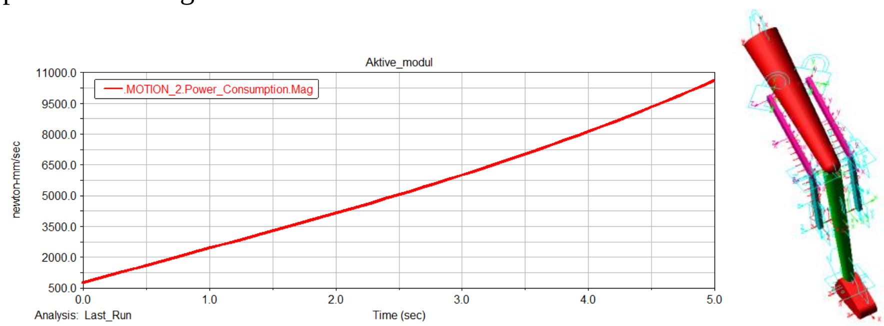

Fig. 5. Power consumption of actuators without considering the flexibility

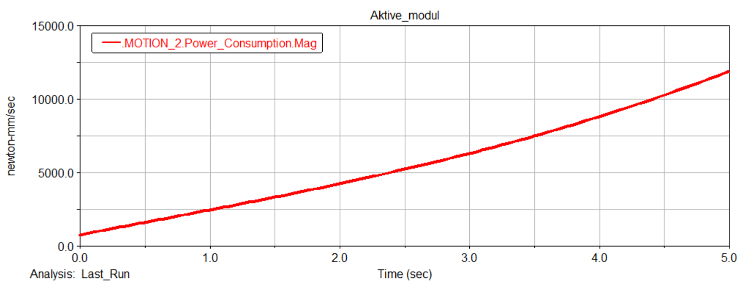

Fig. 6. Power consumption of actuators with considering the flexibility

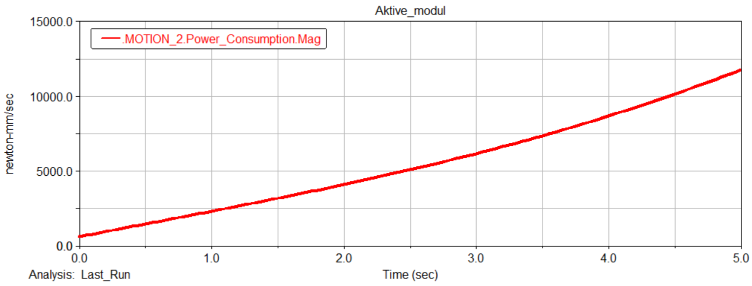

Fig. 7. Power consumption of actuators with considering the activity of the rectus femoris muscle

Conclusions

1. The actuator system design for the rehabilitation device was performed, and the actuator's load carrying ability was checked by the Euler formula. The relation between the mechanical stress and the electrical parameters of the actuator were found. The mechanical stress of each actuator must be greater than $1.3 \cdot 10^7$ *Pa.*

2. The dynamic modeling of the rehabilitation device, considering the links' and joints' elasticity was performed. Natural muscle forces were also included in the equations of motion for the purpose of restoring the impaired balance caused by diseases of motor neurons. The simulation was performed with the program ADAMS, the values of power consumption were defined in three cases: considering the rigid model; considering the flexible model; considering the activity of the rectus femoris muscle. The result showed that the value of the necessary power consumption increased by about 12% in case of a flexible model, the value of the necessary power consumption decreased by about 1.3% in case of considering the activity of the rectus femoris muscle.

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ՄԱՐԴՈՒ ՇԱՐԺՈՂԱԿԱՆ ԳՈՐԾԱՌՈՒՅԹՆԵՐԸ ՎԵՐԱԿԱՆԳՆՈՂ ՍԱՐՔԻ ՀԱՄԱՊԻՏԱՆԻ ԱԿՏԻՎ ՄՈԴՈՒԼԻ ՄՈԴԵԼԱՎՈՐՈՒՄ

Ն.Բ. Զաքարյան

Կատարվել է մարդու շարժողական նեյրոնների հիվանդությունների հետևանքով հենաշարժողական ապարատի կորսված գործառույթները վերականգնող սարքի ակտիվ, համապիտանի մոդուլի կառուցվածքային և դինամիկական մոդելավորում՝ պոլիմեր-մետաղական կոմպոզիտային ակտուատորների կիրառմամբ: Ուսումնասիրվել են նաև այդ ակտուատորների մեխանիկական բնութագրերը: Առաջարկվել է վերականգնողական սարքերի դինամիկական մոդելավորման նոր մեթոդ՝ դրանց օղակների և հոդակապերի առաձգական հատկությունների և ակտիվ կմախքային մկանների ազդեցությունների հաշվառմամբ:

Առանցքային բառեր. վերականգնողական սարք, պոլիմեր-մետաղական ակտուատոր, կառուցվածքային և դինամիկական մոդելավորում, օղակների և հոդերի առաձգականություն:

МОДЕЛИРОВАНИЕ УНИВЕРСАЛЬНОГО АКТИВНОГО МОДУЛЯ РЕАБИЛИТАЦИОННОГО УСТРОЙСТВА ДВИГАТЕЛЬНЫХ ФУНКЦИЙ ЧЕЛОВЕКА

Н.Б. Закарян

Выполнено структурное и динамическое моделирование универсального активного модуля реабилитационного устройства с полимерметаллическими композиционными актуаторами для восстановления утраченных функций опорно-двигательного аппарата при нарушениях двигательных нейронов человека. Проведено исследование механических характеристик актуаторов, предложен новый подход для динамического моделирования модулей реабилитационных устройств с учетом упругости их звеньев и суставов, а также влияния активных скелетных мышц.

Ключевые слова: реабилитационное устройство, полимерметаллический актуатор, структурное и динамическое моделирование, упругость звеньев и суставов.