Preprints of the 21st IFAC World Congress (Virtual)
Berlin, Germany, July 12-17, 2020

On the Influence of the Moment of Inertia on the Mechatronic Drive Control Quality within the Exoskeleton

Stebulyanin M.M.*. Ermolov I.L., Sukhanov A.N.****

** Moscow State Technological University STANKIN, 127055, Vadkovskiy per. 3a, Moscow, Russia (e-mail: mmsteb@rambler.ru).*
*** Laboratory of Robotics and Mechatronics, Ishlinsky Institute for Problems in Mechanics RAS, 119526 Prospect Vernadskogo 101-1, Moscow, Russia (e-mail: sukhanov-artyom@yandex.ru)*

Abstract: The paper considers the interaction between human arm and electromechanical system of the exoskeleton device. Authors proposed the mathematical model of human-exoskeleton interaction on the example of an EMG controlled mechatronic DC-motor. Features of the torque generation with the mechatronic drive of the exoskeleton under the conditions of desynchronization of the speeds of the operator's arm and the exoskeleton link are shown. The exoskeleton motion simulation results with different moments of inertia of the drive system are presented.

Keywords: Human operator support, Design, Modelling and analysis of HMS, Assistive technology and rehabilitation engineering, Telepresence, Telerobotics, Mechatronics, Motion Control Systems, Dynamics and control.

1. INTRODUCTION

One of the practical purposes of the researches in the field of wearable robotics is the development of different exoskeleton systems for moving large and heavy objects, taking into account the imposed requirements for positioning accuracy. Complex, monotonous actions that require, in addition, accuracy in various technological operations can be carried out by the active exoskeleton device. At the same time, the operator's participation in decision-making when manipulating a heavy object allows to realize the flexibility of technological processes, while increasing the efficiency and productivity of operation.

The specificity of the exoskeleton usage in the conditions of bilateral influence of the exoskeleton's links and operator's limbs causes setting special requirements for control, accuracy of operations and safety of the movement. The movements of the exoskeleton links should ensure the safety for the operator's hand and at the same time the control system should implement the desired movement in response to the disturbing effect, represented as a function of the state of the human muscles activity. This is especially true for the development of loading exoskeleton devices operating in a wide range of forces. Such systems are characterized by the presence of external disturbances that exceed human capabilities [Kazerooni, (1993)]. When developing an exoskeleton applicable for loading operations, it is necessary to follow the restrictions on the control.

Unlike the traditional control technique used in the bilateral control systems, which involves control panel device with a force reflection system, the use of electrical activity of the muscles allows to make the system "operator-exoskeleton", in which the operator is directly involved in control. But in some cases of emergency the operator may not disconnect from the control system immediately, which may result in injury. In addition, during operation the operator should sense external forces acting on the exoskeleton links, thus the mechanical structure of the exoskeleton must be equipped with reversible gears. Self-braking transmissions in the mechanical structure are not able to give adequate information about the load torque directly through the exoskeleton's link to the operator. The use of an exoskeleton with mechanical decoupling from the operator will complicate the design in technical terms.

In our work we study the control system that receives the task for the drives of the exoskeleton from the muscular activity of a human [De Luca, (2002)]. This type of control is intuitive for operator when performing manipulative operations with different objects.

The problems of human interaction with a robotic system that repeats the biomechanics of the operator's limbs require careful study due to the presence of a number of restrictions on kinematics, control and modes of operation. One of the important aspects that affect the dynamics of the human-machine system "operator-exoskeleton" is taking into account both the inertia of the links and actuators of the exoskeleton itself, and the inertia of the operator's limbs when manipulating the object.

In our previous works [Gradetsky et al (2017,2018a,b)] the simplified system "operator-exoskeleton" has been discussed and analysed. There has been made an assumption about the rigid connection presented between the master system, represented by the operator's arm, and the slave system of the exoskeleton, represented by the mechatronic drive and the link equipped with handle. The analytical description of such system with a gap between the operator's arm and the handle, which is referred to the technical implementation of such a human-machine system, has been also proposed. This article discusses the features of the interaction of elements in such a system, namely, the influence of the mechatronic drive response on the change in the control signal, which in turn is a reaction to the changed state of the system.

2. MATHEMATICAL MODEL

The main idea of this study was to organize the work of two different subsystems taking into account their influence on each other. In this paper we consider the exoskeleton device controlled via electrical activity of the operator's muscles. The mathematical model of the muscle, which is used in this work, was presented earlier by Alonso et al, (2012) and Thelen (2003). The desired force that a muscle creates $F_{des}(t)$ is expressed as follows [Zajac et al (1989a,b)]:

$$F_{des}(t) = a(t) F_{\max} f_{FV}(V_a) f_{Fl}(l_a) \tag{1}$$

Here $f_{FV}(V_a)$ represents the functional dependence of the contraction force from the velocity of muscle contraction, $f_{Fl}(l_a)$ is the dependence of contraction force from the current length of the muscle [Romero et al (2016), Sancho-Bru et al (2011)]. The level of activation $a(t)$ is a value determined by the ratio of the contraction effort $F_{des}(t)$ that is necessary to create via muscle, to its maximum possible effort, which could be developed at the current parameters of the length $l_a(t)$ and contraction velocity $V_a(t)$. Based on equation (1), the level of muscle activation $a(t)$ is determined as follows:

$$a(t) = \frac{F_{des}(t)}{F_{\max} f_{FV}(V_a) f_{Fl}(l_a)} \tag{2}$$

In our simulation the level of muscle activation is the complex variable depends on all these parameters [Komantsev et al (2001)].
Each skeletal muscle has motor units. Each motor unit is represented with externally excited cell surrounded by a solution of potassium and sodium ions. The motor units are connected to each other sequentially in threads, and the threads themselves are collected in bundles [Zatsiorsky et al (2012)]. A nerve impulse generated by the human brain goes to the bundles of motor units to stimulate them. In response to stimulation, an excitation wave passes through the membranes of the cells of the motor units, changing the electrical potential of the cell membranes.
The value of the activation level $a(t) = [0..1]$ is formed by the central nervous system of the operator on the basis of the planned action and the current tactile and visual information from the feedback channels. It shows the intensity of the muscle contraction, that is, the proportion of all involved motor units per time.
Thus, the operator in the mathematical model can be represented as a task scheduler for moving the desired limb, controlling this movement via the result of the work of the drive system of the exoskeleton. In other words, the operator tries to form the desired movement of the object placed in the hand, while forming an effort on the muscle $F_{des}(t)$, based on the magnitude of the misalignment between the desired angle $\alpha_{des}(t)$ in the elbow and the current angle $\alpha_{real}(t)$ reproduced by the mechatronic drive in the elbow joint of the exoskeleton.

$$F_{des}(t) = K_f(\alpha_{des}(t) - \alpha_{real}(t)) \tag{3}$$

Here K_f is the proportional factor (reaction of the operator).

Each muscle, as an element of a biomechanical system, has a number of current parameters: the current non-zero length $l_a(t)$, the current velocity of contraction $V_a(t)$ and the current developed force of contraction. In addition to the current parameters, the muscle has its own boundary conditions. They include the maximum length $l_{\max}$, the minimum length $l_{\min}$ when contracting, the optimal length l_O [Silva (2003,), Latash (2011)], and the maximum possible contraction force $F_{\max}(l_a(t), V_a(t))$ that a muscle can create for the current length and contraction rate.

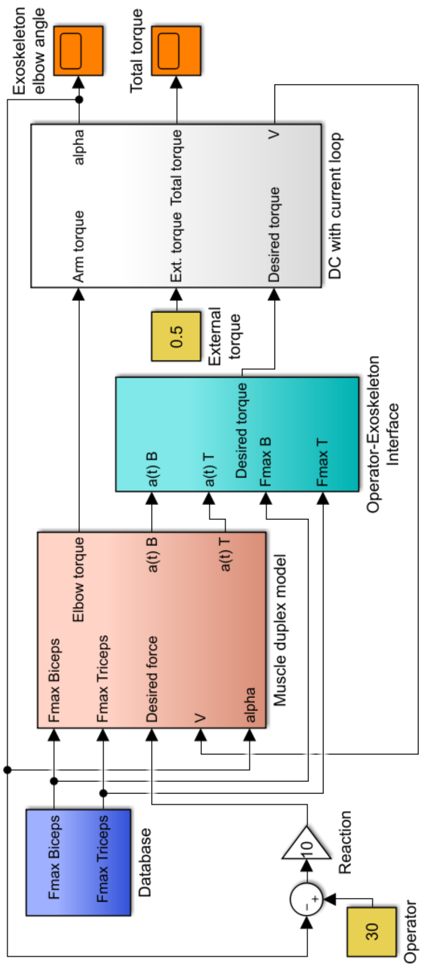

Fig. 1. General control scheme of interaction in "operator-exoskeleton" system

Figure 1 shows the general control scheme that represents interaction within operator-exoskeleton system. The angle and inertia of the operator are simulated, and these are input into the task scheduler. In this scheme, there is the mathematical model of a muscle duplex, an interface block and a model of a DC motor with current feedback loop. In this scheme feedback is necessary for the correct operation of the entire system. Visual position feedback allows the operator adjusting the task to form an effort with his muscles working in pair. Velocity feedback allows calculating the necessary parameters in the muscle model. Also in this scheme there is an external moment, constantly acting on the exoskeleton. It represents the influence of the object.

The scheme also includes a database that contains numerical values of maximum biceps brachii and triceps brachii forces obtained from isotonic and isometric measurements. Figure 2 shows a schematic representation of the muscle duplex that contains mathematical models of antagonist muscles, biceps brachii and triceps brachii.

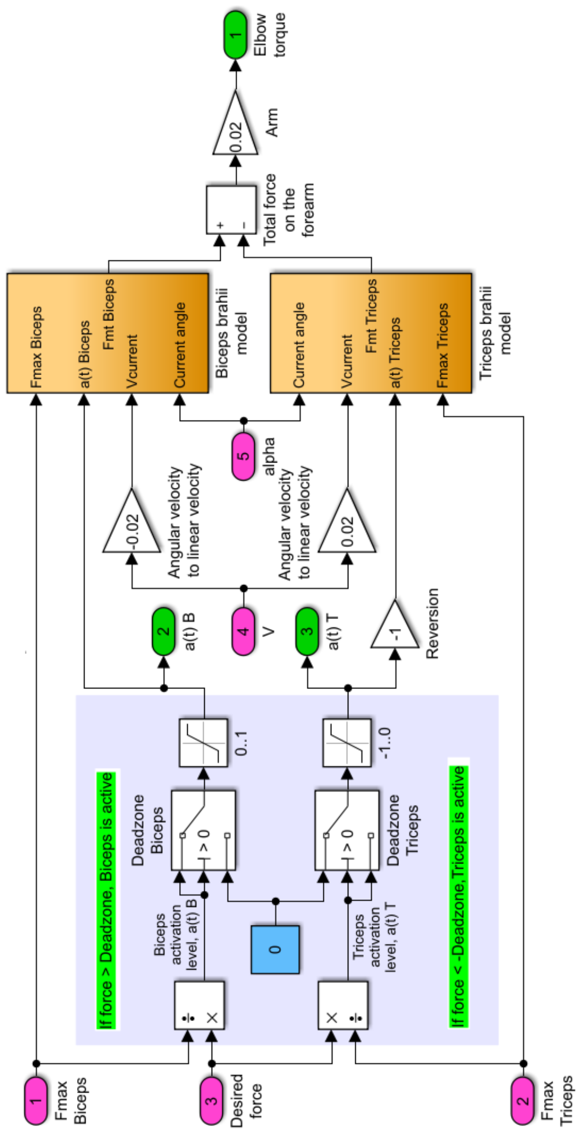

Fig. 2. Muscle duplex model (here the Deadzone is 0)

The formed value of the desired effort should be processed via two channels – the technical channel and the biological one (Fig.2). The technical data channel is necessary to get the activation level obtained by the electromyogram (EMG) sensors, which are given in the interface block. The advantage of applying electromyography to control is concluded in the fact that the obtained signal can be processed before contraction of the muscle fiber completes. The biological information channel contains the description of the of muscle fibers' contraction processes [DiCicco (2004), Zaichenko (2001)]. These processes allow moving the upper limb of the operator, forming a torque in the elbow joint of the arm. This scheme takes into account the switching of the muscles activity to simulate oscillating movements.

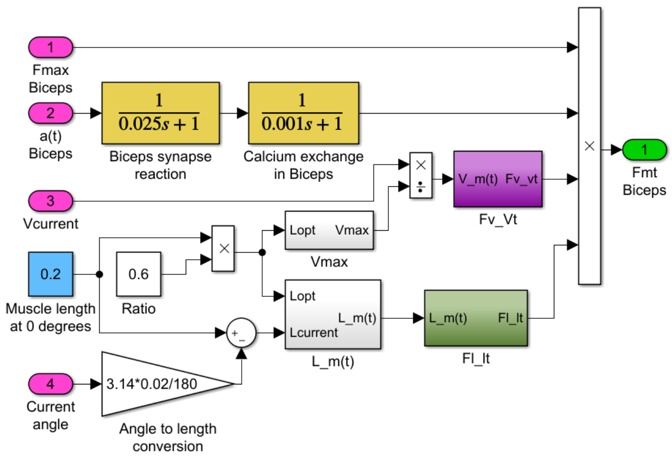

Fig. 3. Biceps brachii model

Figure 3 shows a mathematical model of the operator's biceps brachii. This model describes the dependence of the contraction force from the velocity of the muscle contraction $f_{FV}(V_a)$, as well as the contraction force from the current muscle length $f_{Fl}(l_a)$. In this model the dependence of the contraction force from the current muscle length is determined as follows [Alonso (2012)]:

$$f_{Fl}(l_a) = \begin{cases} 0, & if \ \frac{l_a}{l_o} \le 1 \\ e^{\frac{-\left(\frac{l_a}{l_o}-1\right)^2}{\gamma}}, & if \ \frac{l_a}{l_o} > 1 \end{cases} \quad (4)$$

Here l_a is the current muscle length, calculated from the geometry of the arm, l_o is the optimal muscle length (length at which the muscle can create maximum force).

The force-velocity factor $f_{FV}(V_a)$ depends on the current contraction velocity of the muscle and is defined as [Haeufle (2010)]:

$$f_{FV}(V_a)=\begin{cases} 0, & if\ \dfrac{V_a\tau_c}{l_o}\le -1 \\ \dfrac{l_o k_{CE1}+V_a\tau_c k_{CE1}}{k_{CE1}l_o - V_a\tau_c}, & if\ -1<\dfrac{V_a\tau_c}{l_o}\le 0 \\ \dfrac{l_o k_{CE2}+f_V^{\max}V_a\tau_c}{k_{CE2}l_o + V_a\tau_c}, & if\ \dfrac{V_a\tau_c}{l_o}>0 \end{cases} \quad (5)$$

Here τ_c is the reaction time of the muscle fibres, γ, k_{CE1}, k_{CE2} and $f_V^{\max}$ are factors, which calculated empirically. The following numerical values of these factors obtained in [Thelen (2003)] were used in the simulation process:

$$\tau_c = 0.1, \gamma = 0.45, k_{CE1} = 0.25, k_{CE2} = 0.06, f_V^{\max} = 1.6$$

The model also takes into account the dynamic processes occurring in the muscle body, namely the reaction of synapses to the control signal coming from the central nervous system, as well as the reaction of calcium metabolism occurring in the membranes of muscle fiber cells, which leads to the ion saturation of the space between actin and myosin proteins and the subsequent contraction of motor units. The triceps brachii model looks similarly. It has its own length settings and coefficients [Pinchuk (2002)].

The current biopotential data is processed in the interface block (Fig.4), which is represented by inertial first order transfer functions, modeling the processes occurring in the processing of biopotential data with EMG sensors. This unit allows obtaining the triceps brachii and biceps brachii contraction forces via the current activation level and information taken from the database. It calculates torque in the operator's elbow joint and uses it as task for the control system.

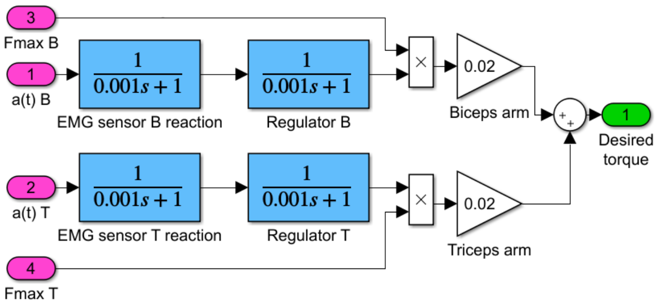

Fig. 4. The interface block

This interface block simulates EMG processing unit as a regulator that issues a control signal to the input of the exoskeleton drive. It has configurable parameters of integration time and ratio factor.

The formed movement velocities of the operator's arm and the exoskeleton links may differ, so there are situations when the operator's arm can either help to move the link of the exoskeleton or interfere with it. This situation was considered when simulating a DC-motor as part of the elbow joint of the exoskeleton device in the form of a variable inertia block structure formed by comparing the autonomous speeds of the operator's arm and the exoskeleton link (Fig.5).

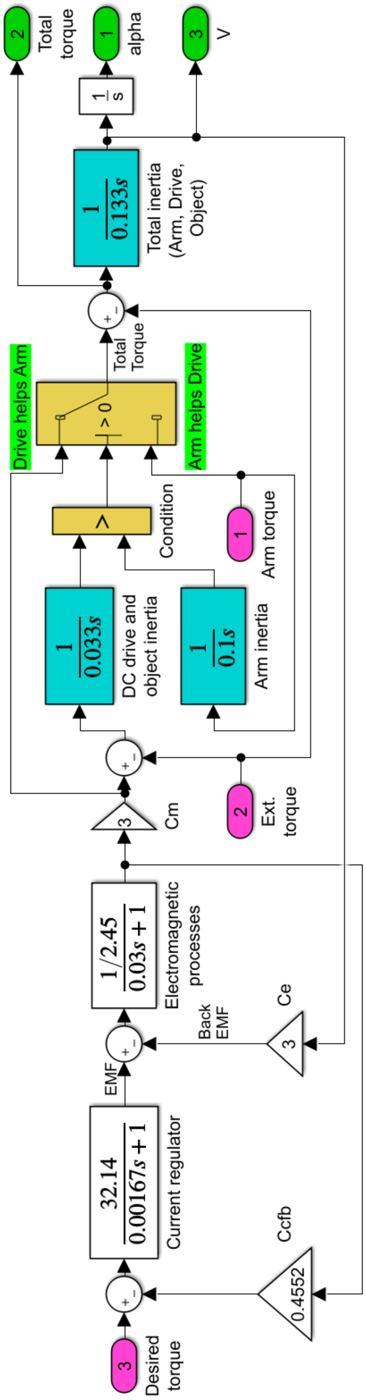

Fig. 5. DC-motor with current feedback loop

The main feedback in the system uses the motor current. Back EMF is calculated at the output of the total inertia unit. The combination of control inputs is carried out using torques of operator's arm and external torque from the object.

3. SIMULATION

The simulation of the system was carried out in the MATLAB Simulink software. Figure 6 shows the results of simulation of the torque engine during the movement of the exoskeleton link in conditions of desynchronization with the movement of the operator's hand. The desynchronization was achieved by different model response times in the system (see fig.3 and 4).

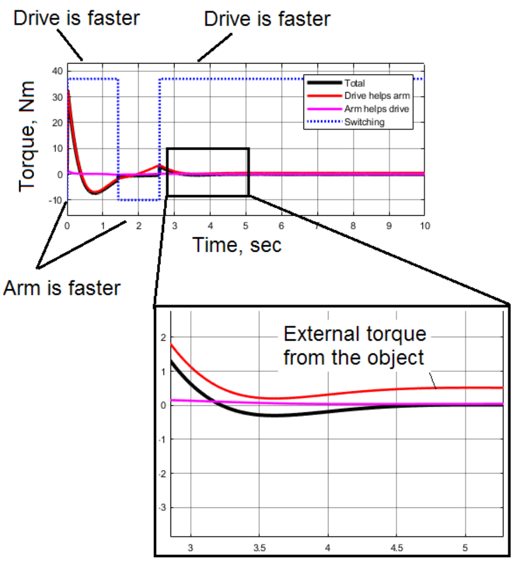

Fig. 6. Torques in the simulation (motor inertia here is $0.033 kg*m^2$, arm inertia is $0.1 kg*m^2$)

Figure 6 shows a simulation of the process of moving a load attached to the handle of the exoskeleton. At the initial time moment, the torque value of the motor is maximum. Due to the presence of inertia, the link still continues to move after reaching the set value of the angle at the elbow (30 degrees). Seeing this, the operator strains the triceps brachii to make the motor run in reverse (switching). Thus, in a couple of iterations, the system enters the steady operation mode, which is expressed in holding a load of a certain mass.

Figure 7 shows the effect of influence of the moment of inertia of the exoskeleton drive on the quality of control with fixed moment of inertia of the operator's arm.

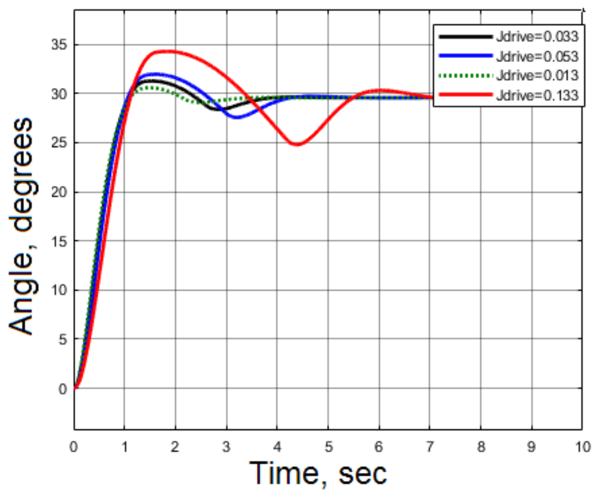

Fig. 7. Transient processes in simulation with inertia variation

Thus, with a decrease in the moment of inertia of the drive compared to the moment of inertia of the human arm, there is a tendency of reducing the overshoot and reducing the transition time, while the increasing in the moment of inertia of the exoskeleton drive compared to the moment of inertia of the human arm reduces the characteristics of the control quality.

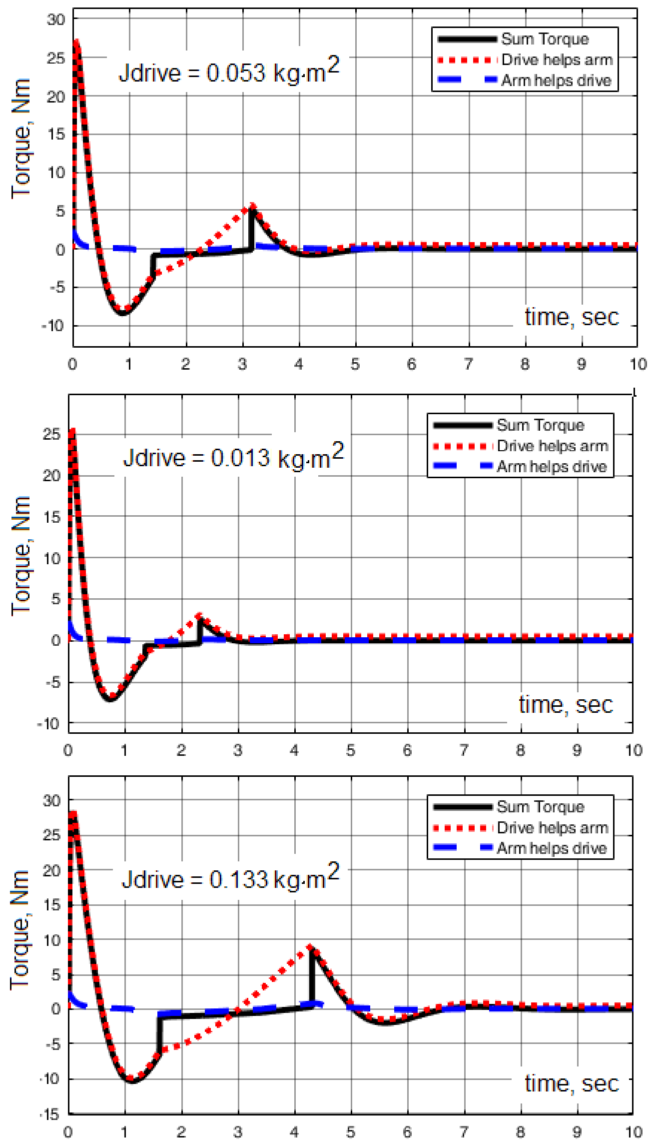

Fig. 8. Torques in the simulation with variation of motor inertia (arm inertia is still $0.1 kg*m^2$)

As one can see from the figures 7 and 8, the lighter link of the exoskeleton allows reaching the desired angle faster. While the heavier link of the exoskeleton requires more time for this operation.

This behaviour in the simulation can be explained by the fact that a more inertial drive requires more time for acceleration and braking, which in turn can reduce the safety of the exoskeleton operating. This simulation model makes it possible to estimate the contribution of the moment of inertia and other parameters in the model to overshoot during the movement of the exoskeleton link. Restrictions imposed by the physiology of human joints should set limits for the movement of exoskeleton links under different types of control (fast or slow)

6. CONCLUSIONS

In this paper, a mathematical model of the electromechanical drive of the exoskeleton with biofeedback was proposed. It represents the possibility of desynchronization of the operator's arm movement and the movement of EMG controlled exoskeleton. The simulation showed that there may occur such situations in which the operator can either help to move the exoskeleton link, reducing the resistance torque, or to interfere with the movement, creating an additional disturbing torque. The dependence between the moment of inertia of the mechatronic drive of the exoskeleton and the quality of its control was also revealed.

ACKNOWLEDGMENTS

This research was supported by Basic Research Program I.7 "New Developments in Perspective Areas of Energetics, Mechanics and Robotics" of the Presidium of Russian Academy of Sciences and also is supported by the Ministry of Science and Higher Education within the framework of the Russian State Assignment under contract No. AAAA-A17-117021310384-9

REFERENCES

Alonso, J., Romero, F., Pàmies-Vilà, R., Lugrís, U., and Font-Llagunes, J.: A simple approach to estimate muscle forces and orthosis actuation in powered assisted walking of spinal cordinjured subjects, *Multibody Syst. Dyn.*, 28, 109–124, 2012.

De Luca C. J. *Surface Electromyography: Detection and Recording* / Carlo J. De Luca, 2002 by DelSys Incorporated, 2002. - P.10.

DiCicco, M. Comparison of Control Strategies for an EMG Controlled Orthotic Exoskeleton for the Hand / Matthew DiCicco, Lenny Lucas, Yoky Matsuoka // *Proceedings of the 2004 IEEE International Conference on Robotics and Automation, New Orleans*, LA, April 2004. – pp. 1622-1627.

Gradetsky, V.G., Ermolov I.L., Knyazkov M.M., and Sukhanov A.N.. Generalized approach to bilateral control for EMG driven exoskeleton. In Conf.Volume 113, 2017, *12th International Scientific-Technical Conference on Electromechanics and Robotics*, volume 113 of Zavalishin's Readings, pages 1–5, 2017.

Gradetsky, V.G., Ermolov I.L., Knyazkov M.M., Semenov, E.A., and Sukhanov A.N. The dynamic model of operator-exoskeleton interaction. In *Interactive Collaborative Robotics, proc. of the 3rd International Conference ICR 2018*, pages 52–59. Springer Nature Switzerland AG 2018 Printforce, the Netherlands, 2018.

Gradetsky, V.G., Ermolov I.L., Knyazkov M.M., Semenov, E.A., and Sukhanov A.N. The influence of various factors on the control of the exoskeleton. In *Interactive Collaborative Robotics, proc. of the 3rd International Conference ICR 2018*, pages 60–70. Springer Nature Switzerland AG 2018 Printforce, the Netherlands, 2018.

Haeufle, D. F. B., S. Grimmer, A. Seyfarth a 2010 The role of intrinsic muscle properties for stable hopping—stability is achieved by the force–velocity relation. *Bioinspir. Biomim. 5*, 016004. doi:10.1088/1748-3182/5/1/016004

Kazerooni, H. *Human extenders* / H. Kazerooni, J. Guo // ASME J. Dyn. Syst. Meas. Contr. 115(2B), 1993. - pp. 281–289.

Komantsev V.N., Zabolotnykh V.A. Metodicheskie osnovy klinicheskoi elektroneiromiografii: ruk. dlia vrachei [Methodological basis of clinical electroneuromyography: guide for physicians]. SPb., Lan', 2001, 349 p. (In Russian)

Latash, M. L. (2011). Anticipatory control of Voluntary action - Merging the ideas of equilibrium-point control and synergic control. in Motor Control - *Theories, experiments and applications*, pp. 3–26.

Pinchuk D.Iu., Dudin M.G. Biologicheskaia obratnaia sviaz' po elektromiogramme v nevrologii i ortopedii: spravochnoe ruk. [Biological feedback by electrical myogram in neurology and orthopaedics: reference manual]. SPb., Chelovek, 2002, 120 p. (In Russian)

Romero, F., & Alonso, F. J. (2016). A comparison among different Hill-type contraction dynamics formulations for muscle force estimation. *Mechanical Sciences, 7(1)*, 19–29. https://doi.org/10.5194/ms-7-19-2016

Sancho-Bru, J.L., Antonio Pérez-González, Mora, M.C., León, B.E., Vergara, M., Iserte, J.L., Rodríguez-Cervantes, P.J., Morales, A.: Towards a realistic and self-contained biomechanical model of the hand. *Theoretical biomechanics* pp. 212–240 (2011), http://www.intechopen.com/articles/show/title/towards-a-realistic-and-self-contained-biomechanical-model-of-the-hand

Silva, M.: Human motion analysis using multibody dynamics and optimisation tools, PhD thesis, Instituto Superior Técnico, Unversidade Técnica de Lisboa, Lisbon, Portugal, 2003.

Thelen, D.: Adjustment of muscle mechanics model parameters to simulate dynamic contractions in older adults, J. Biomech. Eng.-T. ASME, 125, 70–77, 2003.

Zaichenko K.V., Zharinov O.O., Kulin A.N., Kulygina L.A., Orlov A.P.; Ed. Zaichenko K.V. S"em i obrabotka bioelektricheskikh signalov: ucheb. posobie [The removal and processing of bioelectric signals: manual]. SPb., 2001, 95 p. (In Russian)

Zajac, F. E. and Gordon, M. E., "Determining muscle's force and action in multi-articular movement," *Exerc Sport Sci Rev*, vol. 17, pp. 187-230, 1989.

Zajac, F. E (1989) Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control. *CRC Critical Reviews in Biomedical Engineering* 17, 359-411.

Zatsiorsky, V. and Prilutsky, B.: *Biomechanics of skeletal muscles, Human Kinetics*, Human Kinetics Publisher, Champaign, USA, 2012.