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DEVELOPMEN AND MODELING OF A BIOMECHATRONIC SYSTEM FOR FOOT RECEPTOR ACTIVATION OF PATIENTS-PARAPLEGICS

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Summary: *A biomechanical approach is adopted to synthesise a mechatronic system for human foot receptor activation. Experimentally found data for the dynamic characteristics of a normal walk are used. The application of the system for the rehabilitation of patients-paraplegics is verified by studying the effect of foot receptor signals on locomotion. Different loading regimes are provided. Parameters subjected to regulation are the simulated walk speed and the maximal value of the foot loading.*

The objective of the work presented in this paper is a derivation of a parametrical model for a definition of the proper laws and control of the drives of the system mechanical module for walk in norm of patients with spinal-brain damages.

1. Introduction

It stands clear that durable training of patients-paraplegics based on clinical investigations performed during the recent years, which undergo external support and passive limb movement on a thread-band implies rehabilitation of their capability to perform primitive walk (Diez V., Colombo G., et al., 1995; Shik M.L, 1995; Diez V., Wirz M., et al., 1998). Besides, the value of pressure on the support area (foot) becomes a critical factor for the initiation of locomotion-like movement (pacing).

Normally, walk is a process where the left foot and the right one undergo rhythmical and shock loading, which is applied in a specific succession. Thus, the activation of the foot receptors is automatically maintained. The receptor signals form a sufficient data flow that serves as a biological feedback. The latter maintain locomotion generation and regulation. In case of spinal cord break as a result of a trauma or disease, a serious functional upset occurs.

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Data flow to the high sections of the nervous system is cut, but the information links with the spinal cord from below and to the break spot are kept. Neuro - physiological studies prove that nerve conductivity is lost in a month after the spinal cord break. The performance of the support - locomotion system is blocked. The patient looses his capability of independent movement and immobilization yields atrophy of his skeleton - muscle system (muscles, bones, joints). Considering this set up, the probability of physical, psychical and social rehabilitation of such a contingent of patients is minimal.

2. Methods

2.1. Task, problems and approach employed for the development of a specialized mechatronic system

The basic task of the study is as follows: considering patient's serious trauma, find a method for the simulation of patient's walk close to the normal one; maintaining data flow from the foot receptors, stimulate the locomotion capabilities of the spinal cord and the functioning of patient's support - locomotion system.

We develop a specialized mechatronic system to fulfil the task and to solve the following problems:

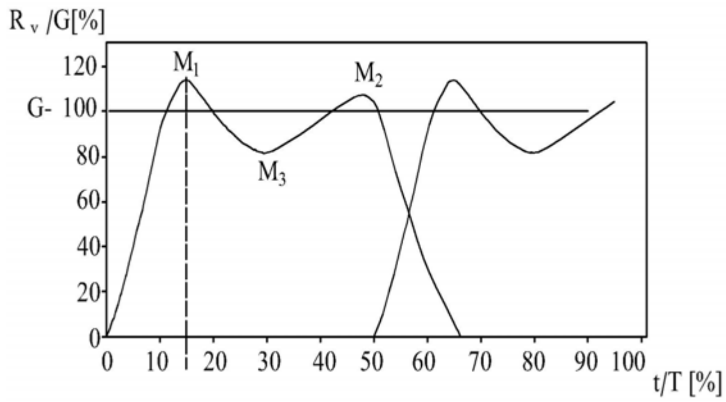
1. Attaining maximal comfort of the system individual use by patients -paraplegics during the earliest possible post - traumatic period;
2. Limitation of the atrophy of patient's support - locomotion system, yet without the performance of locomotion - like movement of the lower limbs;
3. Initiation of data flow from the foot receptors to stimulate the rehabilitation of the locomotion capabilities of the spinal cord;
4. Maintenance of the activation of the nerve ends, located on the feet and corresponding to the internal organs, to stimulate and maintain organ functioning, regardless of patient's immobilization.

We look for a solution of the system mechanics, actuation and control applying a biomechanical approach for the system development that would provide patient's feet loading similar to that of a normal walk. We use as a standard the graph of the vertical support reaction in the single - support stage of walk.

2.2. Dynamics of human foot loading

Biomechanical studies show that one of the most sensitive and informative parameter characterizing locomotion and reflecting the state of the human support – locomotion system is the vertical component of the support reaction occurring during walk (Yanson H.A., 1975; Winter D.A., 1990; Willams & Lissner, s, 1992).

Figure 1 shows the graph of the vertical support reaction of a foot under normal walk.



M_1 –1-st maximum
M_2 –2-nd maximum
M_3 – minimum
G – weight of the investigated man
T – period of the gait
____ right leg
------ left leg

Fig. 1: A graph of the vertical support reaction of a foot under normal walk

Maximum 1 corresponds to the holding period of the support stage – heel support. Maximum 2 corresponds to the push period of the support stage – toe support. The local minimum of the curve corresponds to the shift of the opposite leg, i.e. to minimal loading of the whole supporting foot. The horizontal line of Fig. 1 corresponds to the weight of the investigated man (weight level). All extremums are located significantly above or below the specified level. This illustrates the walk dynamic character and proves that muscle, gravitation and inertia forces effectively operate during walk. The graph allows accounting for the durability of the support stage periods.

We use the graph as a standard when planning patient’s foot loading in accordance with the three periods of the support stage. The curve interpolation is performed at three points (extremums). However, one should reduce the value of the maximal load in favour of the rehabilitation of patients - paraplegics who have been immobilized for a long period of time and considering the occurring atrophy of the skeleton - muscle system.

2.3. Technical solution of the mechatronic system for the simulation of human walk dynamics

2.3.1. System configuration

The general scheme of the specialized mechatronic system is shown in Fig. 2. The system consists of: 1 – module for foot dynamic loading – 2 pieces; 2 – module 2 pieces - controller; 3 – computer for system control (compatible with IBM PC).

The system comprises of two identical modules each one for the left and right foot, respectively. The devices provide foot loading in compliance with the walk rhythm following a given program specified by the computer. Hence, walk with a specified speed is simulated. Periods of single and double support are realized together with the corresponding loading of each foot accounting for the walk temporary characteristic.

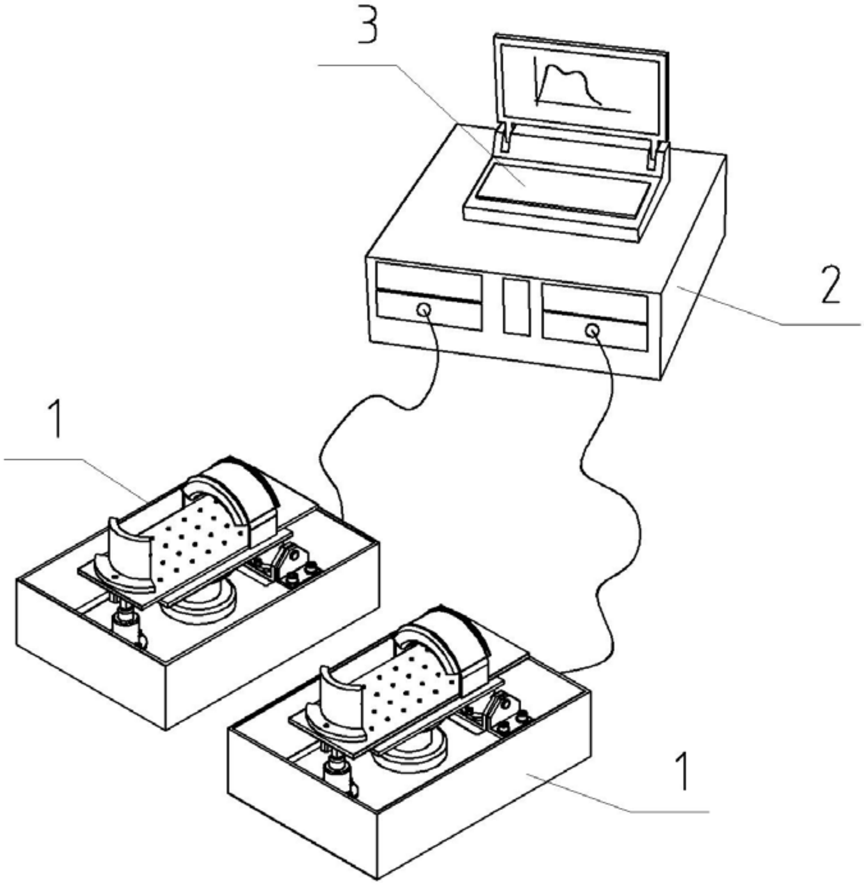


Fig. 2: General scheme of the mechatronic system for the simulation of human walk dynamics

1 – module for foot dynamic loading,

2 – module-controller,

3 – computer for system control.

2.3.2. Structural solution of the module for foot dynamic loading

The structure of the module for foot dynamical loading is shown in Fig. 3. The module structure is described in what follows. A lower immovable plate 1 is used as a structure element and a power transducer 2 is fixed to its centre. Two linear motors 3 are mounted at both ends of the immovable plate 1. One of the motors is firmly fixed while the other one hangs on a hinge on axis 4. A movable plate 7 with apertures is attached to axes 5 of the linear motors by means of hinges 6. Pins 8 with stop rings 9 are inserted in the apertures of the movable plate 7. The lower ends of pins 8 touch plate 10. Plate 10 is mounted on the power transducer 2 which in its turn is fixed to the lower side of the movable plate 7. Foot stops 11 and elastic belts 12 with attached pressing cushions 13 are fixed to movable plate 7.

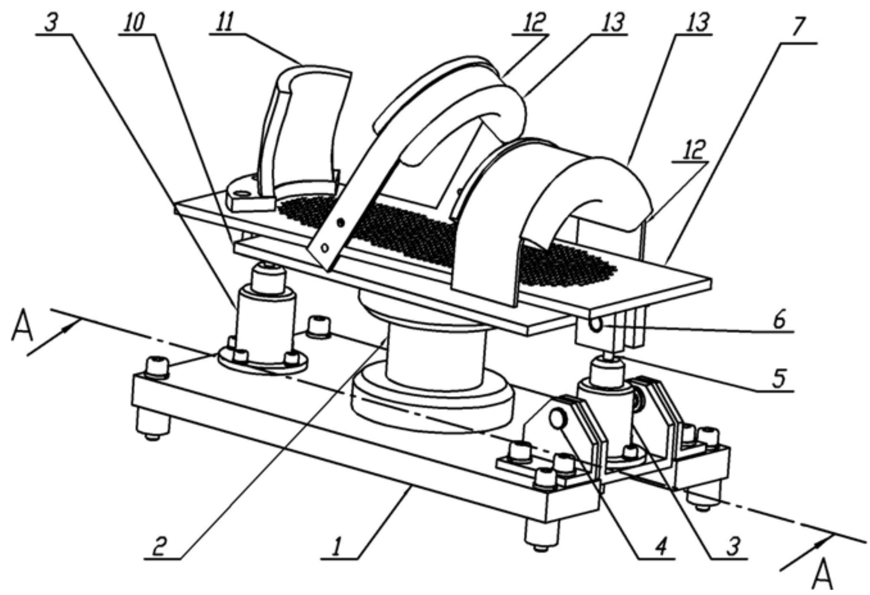


Fig.3: Axonometric view of the module for foot dynamic loading

1 - immovable plate; 2 - power transducer; 3 - linear motor; 4 – axis;

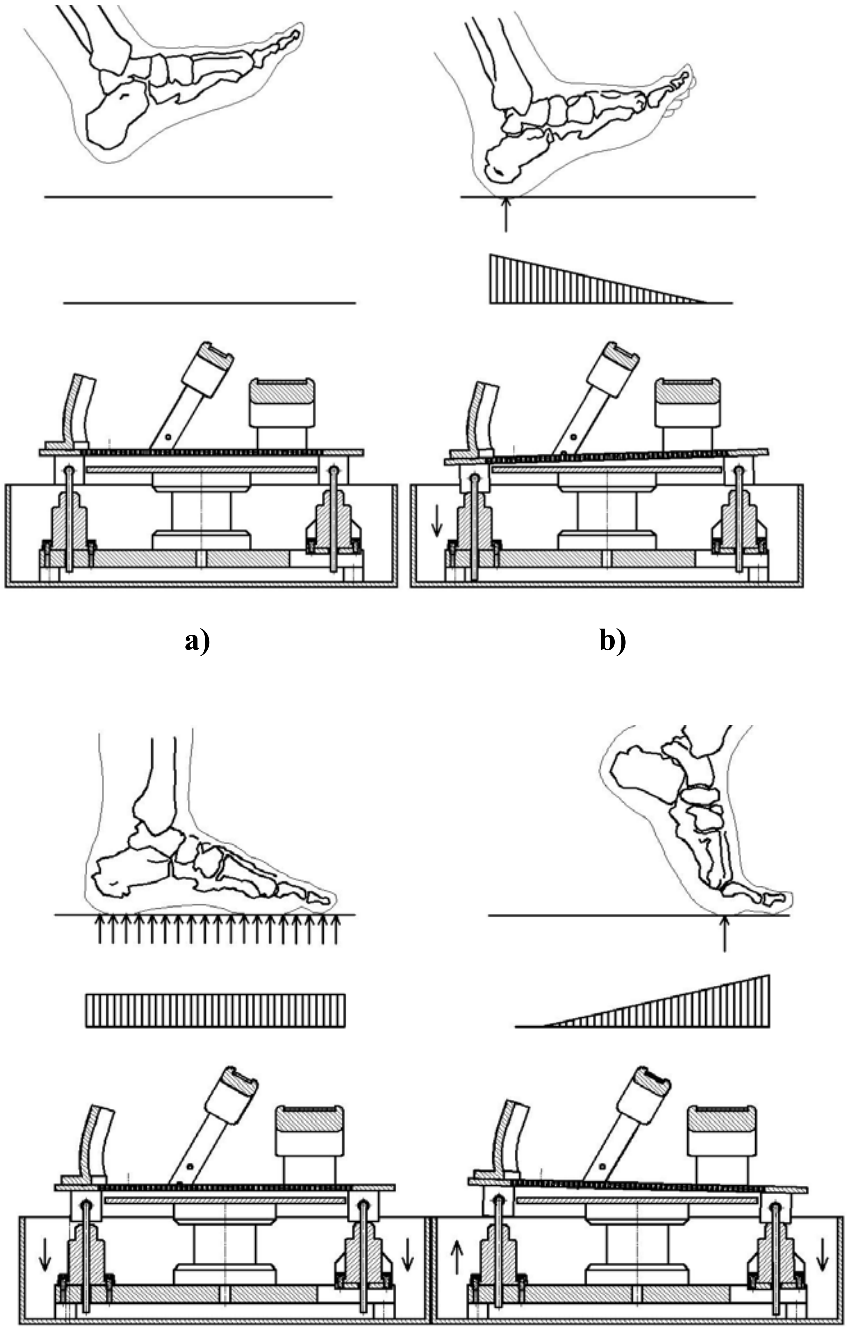
5 – axis; 6 – hinge; 7 - movable plate; 8 – pin; 9 – ring; 10 – plate;

11 - foot stop; 12 - elastic belt; 13 - pressing cushion

The technical solution of the mechatronic system for the simulation of human walk dynamics is presented in details in the publication (Platonov A. et al., 2007).

2.3.3. Functional realization of foot loading in compliance with the walk rhythm

Software sets up walk with a specific period T, while at the same time walk is simulated by the mechanical system – Fig. 4. The two basic stages of walk i.e. walk fly stage (Fig. 4.a) and walk support stage for each leg (Fig. 4. b, c, and d), are simulated during loading. Load transfer from leg to leg is performed. This means that load transfer from a single support of the right leg to a single support of the left leg is performed, passing through the corresponding stages of double support. Foot load for a single support is applied, following the graph of the vertical support reaction plotted for the three stages of the support phase of normal walk (Fig. 1). Figure 4 shows load application for: **a** – fly stage – loading 0 N; support stage: **b** – holding period (heel support), **c** – period of support of the entire foot, **d** – push period (toe support). Loading during heel support (b) is calculated considering M_1-1-st maximum – on the graph of the vertical support reaction of normal walk (Fig. 1).



a) b)

c) d)

Fig. 4: Foot loading during walk rhythm

a- fly stage – loading 0 N; support stage: b, c, d
b – holding period (heel support),

c – period of support on the entire foot,

d – push period (toe support).

Loading during the stage of support of the entire foot (c) is calculated regarding the local minimum of the same graph. Loading during the stage of toe support (d) corresponds to the graph maximum 2. Thus, the graph of the vertical support reaction is approximated at three points. The module thus described performs loading of one of the legs. A similar module is used to load the other leg.

2.4. 3D model of the module for foot dynamic loading

2.4.1. Model basic elements and geometrical dimensions

A simplified 3D model of a device is designed, and its basic dimensions are given in Fig. 5. Generally speaking, the model may be considered as composed of: 1 – immovable elements (stand); D1 - immovable linear motor and fixed to the foundation, with a movable shaft 3; D2 – linear motor with a movable shaft 4, joined to the foundation 1 by a hinge; 2 – a plate, immovable and fixed to the foundation 1; 5 – a movable plate with apertures, joined by a hinge to the movable axes of the linear motors D1 and D2.

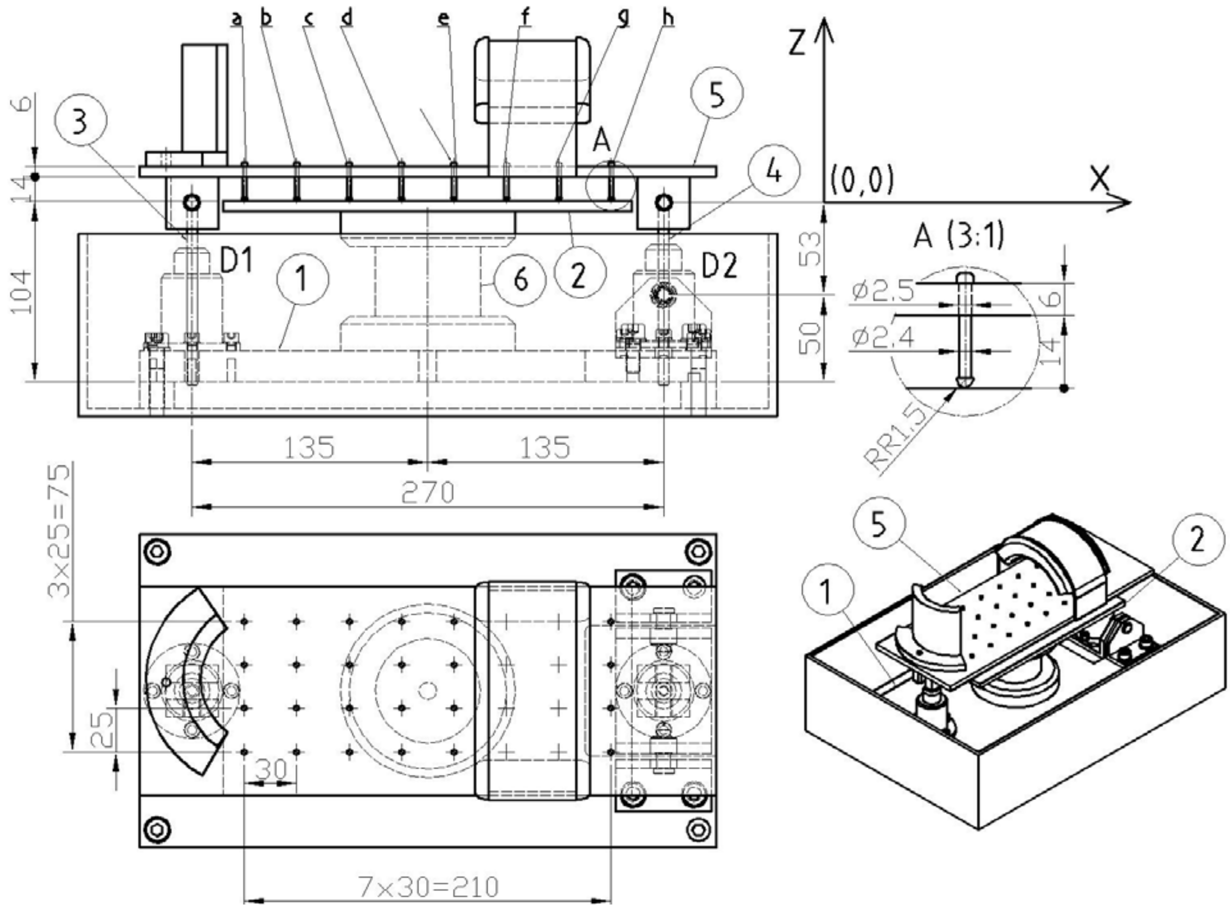


Fig.5. Basic elements and geometrical dimensions of the model

Pins (m x n in number) a, b, c, d, e, f, g and h are mounted in the immovable plate 2 with a windage which allows their motion in axial direction. One pin end contacts with the plane area of the immovable plate 2, and it is formed as a spherical surface. The other end operates on the patient's foot. The step of distribution and the pin dimensions are specified in the figure. Pins a, b, c, d, e, f, g and h shift axially in vertical direction as a result of the displacement of the movable plate 5 with respect to the immovable one 2. For such type of motion, pins change their position with respect to the movable plate 5 and they can operate on a foot which is immovable and placed on the plate. Pins realize contact between the immovable and the movable plate, whereas pressure is applied on the foot and it is recorded by the loading sensor 5.

2.4.2. Modelling of the kinematical joints

The joints between the immovable part 1 (stator) and the movable shaft, as well as those between the stator of motor D2 and the shaft 4 are modelled by means of a linear actuator. The kinematical joints between the movable plate and axes 3 and 4 are revolute ones. Joints between the movable linear motor D2 (item 2) and the foundation 1 are of the same type. The joints between the movable plate and the pins are modelled by means of a rigid joint on a slot in one direction. Joints between the pins and the immovable plate are of the type "spherical joint on plate". Thus, we have designed a virtual 3D model of the mechanical module of the biomechatronic system, which is shown in Fig. 6.

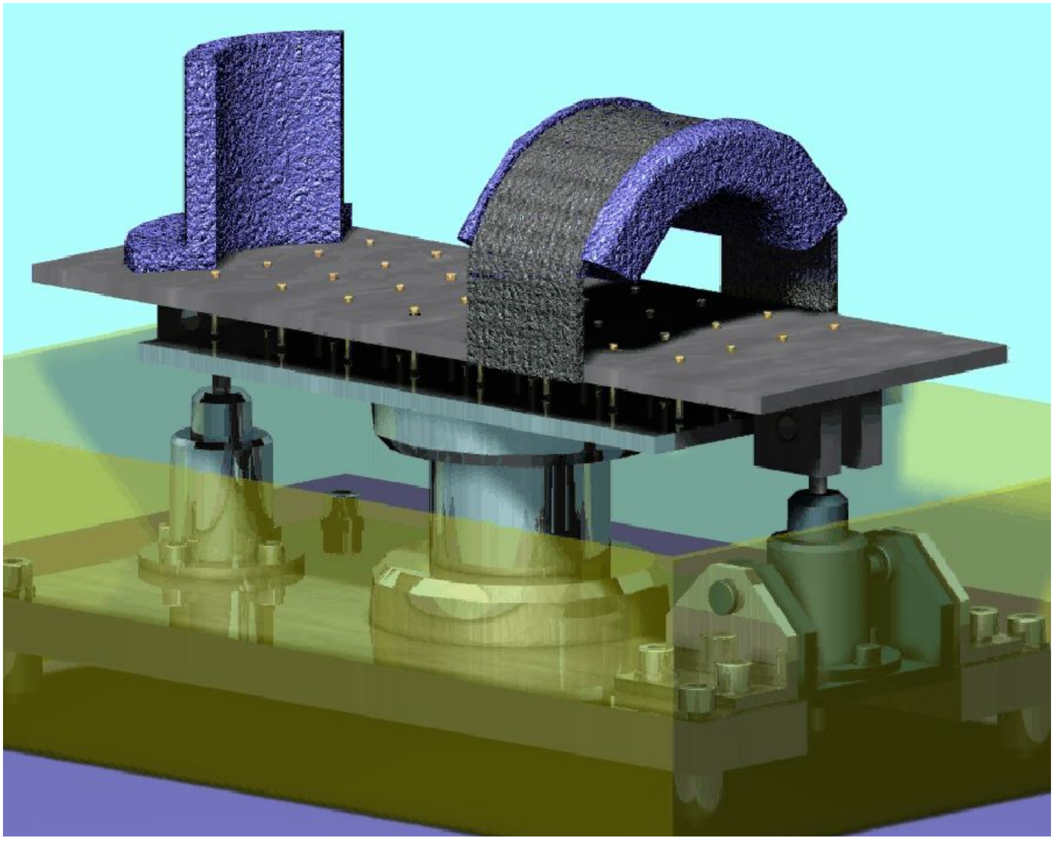


Fig.6: **3D** model of the mechanical module of the biomechatronic system.

2.5 Walk simulation on the 3D model

We formulate the loading task during the three stages of the single support phase of walking, using the dynamical characteristic shown in Fig.1, presented in Fig.7 below:

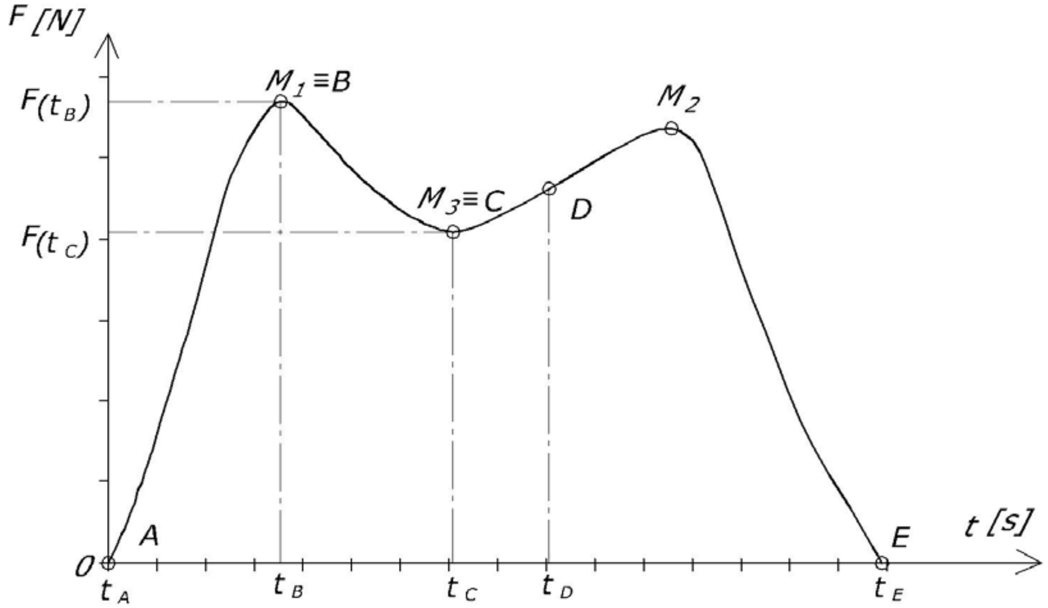


Fig.7: Desired pin displacement set up as function of time

2.5.1. Restricting conditions

1.**W**e assume that motors apply loading in three stages, too, in order to model the three phases of walking:

a. Motor D1 operates during phase 1;
b. Motors D1 and D2 operate during phase 2;
c. Motor D2 operates during phase 3.
d. The duration of phases 1, 2 and 3 (t_B -t_A , $t_D - t_B$, $t_D - t_E$) is shown in Fig. 7, respectively.

2. During phase 2, the first motor D1 moves with constant velocity along axis **Z** until reaching the initial zero position.
3. The displacements of both motors along axis **Z** should be equal to each other at point **C** of the local minimum M_3=$F(t_C)$.
4. We prescribe maximal admissible deformation of the soft tissues under pressure exercised by the pins equal to 7 mm in order to avoid injuring of patient's foot.
5. We assume that foot loading is proportional to the sum of the displacements of all pins along axis **Z.**
6. We must generate a law of linear displacement of motor shafts, based on those restrictions and the displacement should be a function of time. Thus, we could attain foot loading corresponding to the loading task *F(t)* as prescribed in Fig.7.

2.5.2. Synthesis of laws for control of the motors of the walk stimulation module

It follows that loading must be generated during each of the three walking phases. Distribute pins that act on foot into *n* rows. Assume that each row has *m* pins.

1. Considering phase 1 with duration from t_A to t_B, motor D1 should perform displacement Δ1(t), t∈[t_A , t_B] in direction *Z*, and that displacement should yield foot loading ranging from 0 to $M_1=F(t_B)$ in compliance with curve AB. It follows from Fig. 4.b that for each row of pins displacement Δ1(t) of motor D1 would yield total pin displacement of the corresponding order $0.5m\Delta 1(t)$. Hence, the total displacement of all pins will be:

$$0.5mn\Delta 1(t) \tag{1}$$

It follows from (1) that considering displacement Δ1(t), t ∈[t_A , t_B], and foot loading will have the value:

$$F(t) = 0.5kmn\Delta 1(t), \tag{2}$$

where *k* is a coefficient of proportionality between the sum of the pin displacements and the foot loading force. Coefficient *k* is calculated from condition (4) and equation (2) for $\Delta 1(t_B) = 7mm$ at point ***B*** of the local maximum $M_1 = F(t_B)$:

$$k = \frac{2F(t_B)}{7mn}. \tag{3}$$

Then, the law of control of motor D1 will read:

$$\Delta 1(t) = \frac{2F(t)}{kmn} \tag{4}$$

We assume during phase 2 with duration from t_B to t_D (following condition (2)) that control of motor D1 will have the form:

$$\Delta 1(t) = \Delta 1(t_B) - a(t - t_B), t \in [t_B, t_D]. \tag{5}$$

Motion of the second motor D2 along axis ***Z*** is maintained at moment t_B and it is realized according to the following law of operation:

$$\Delta 2(t) = \frac{2F(t)}{kmn} - \Delta 1(t). \tag{6}$$

Coefficient *a* is derived from condition (3): $\Delta 1(t_C) = \Delta 2(t_C)$, (Fig.4.c.), at point ***C*** of the local minimum $M_3 = F(t_C)$:

$$a = \frac{2F(t_B) - F(t_C)}{kmn(t_C - t_B)}. \tag{7}$$

The final moment of phase 2 - t_D is defined from the relation $\Delta 1(t_D) = 0$:

$$t_D = t_B + \frac{2F(t_B)(t_C - t_B)}{2F(t_B) - F(t_C)}. \tag{8}$$

Only motor D2 operates during phase 3 following condition (1), (Fig. 4.d.) (with duration from t_D to t_E), and its operation is controlled by the following law:

$$\Delta 2(t) = \frac{2F(t)}{kmn}. \quad (9)$$

2.5.3. Numerical example

Table 1. Control of motors during walking period T (double step – single-support and swing phases)

Input					Motor control	
Interval	loading	Phase	Time	Pressure force	Displacement of motor 1	Displacement of motor 2
			t [s]	F [N]	Δ1 [mm]	Δ2 [mm]
1	Heel	Single-support	0	0	0	0
			0.05	20.55017	-1.9386952	0
			0.1	47.68894	-4.4989568	0
			0.15	69.49408	-6.5560456	0
2	Whole foot		0.2	72.00403	-6.4228528	-0.36998
			0.25	63.37609	-5.1501216	-0.8287552
			0.3	56.78749	-3.8773904	-1.47992
			0.35	53.33632	-2.6046592	-2.4270688
			0.4	55.68939	-1.3467272	-3.9069888
3			0.45	59.92492	0	-5.6532944
			0.5	64.63107	0	-6.0972704
	Toes		0.55	68.8666	0	-6.4968488
			0.6	68.08224	0	-6.4228528
			0.65	47.68894	0	-4.4989568
			0.7	27.29564	0	-2.5750608
			0.75	11.92224	0	-1.1247392
			0.8	0	0	0
4		Swing	0.85	0	0	0
			0.9	0	0	0
			0.95	0	0	0
			1	0	0	0
			1.05	0	0	0
			1.1	0	0	0
			1.15	0	0	0
			1.2	0	0	0
			1.25	0	0	0

We have performed in the present study simulation of the control of a virtual mechatronic device used to load patient's foot, keeping the rhythm of normal walk. The following parameters were used in the study: m = 8 pins and n = 4 rows, (Fig. 5), $F(t_B)$ = 720 N, $F(t_C)$ = 480 N. We found the control impacts – see Table 1, using the algorithm outlined in Sub-paragraph т.2.5.2, where their values are specified. The tabular data are given as laws of control with respect to the position of the corresponding motors used in the model. Those data are approximated by means of cubic splines. Plots of the synthesized control of both motors of the virtual mechatronic module used for walk simulation are shown in Fig.8.

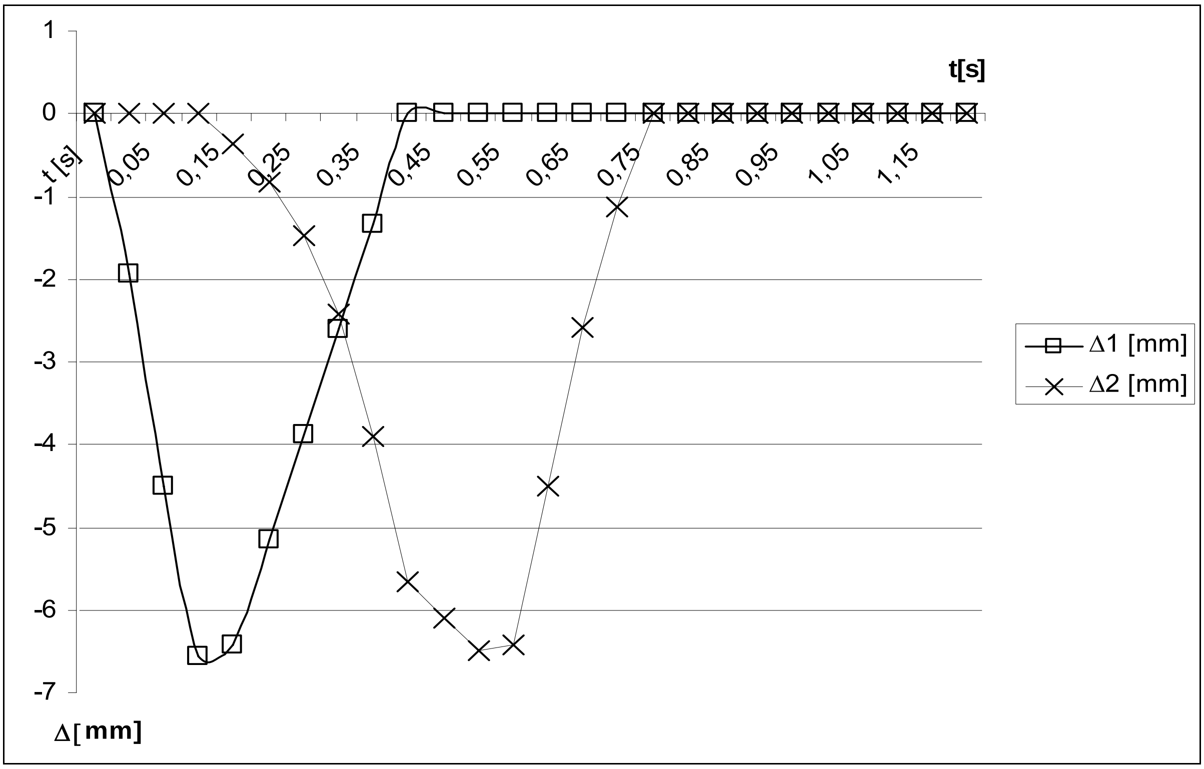
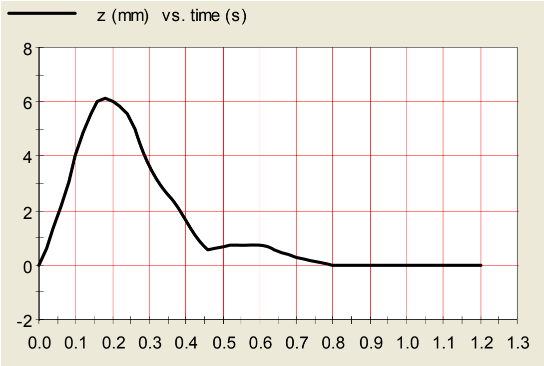


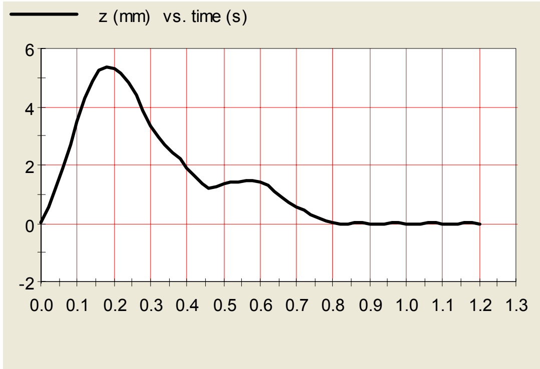
Fig.8. Displacement (Δ1 and Δ2) of the linear motors (D1 and D2) plotted as a function of time; it is used to obtain task F (t) of foot loading.

3. Results and discussion

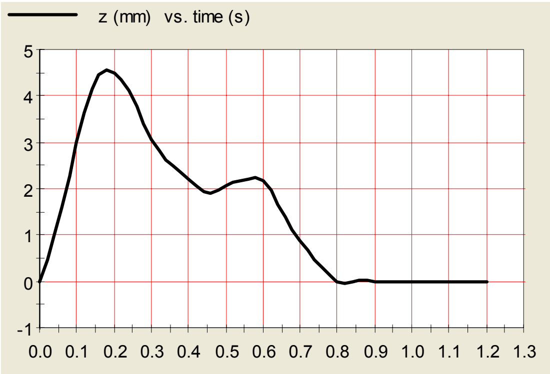
A simulation of motion of linear motors was performed, based on the input parameters and the control signals using specialised software. Plots of the displacement of specific pins belonging to a specific row (the displacement being function of time) are shown in Fig. 9. The plot of the sum of displacements of pins belonging to that row is shown in Fig. 10. It is obvious that the plot of the variation of the total displacement (Fig.10) is similar to the plot *F(t)* of the desired profile function of loading (Fig.7). The virtual 3D model designed herewith enables one to test the different laws of the device control, choose drives and select materials for the fabrication of specific compounds.



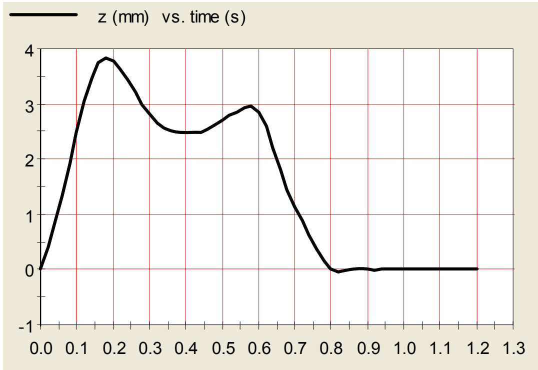
Pin. 1.



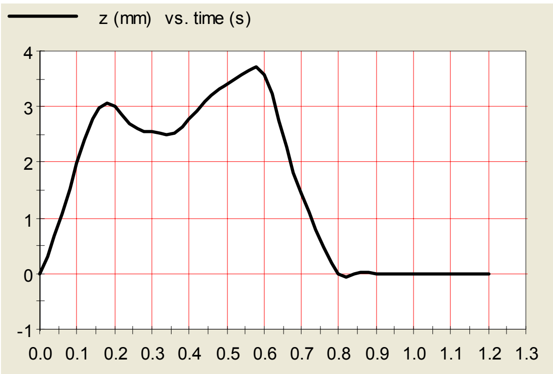
Pin 2.



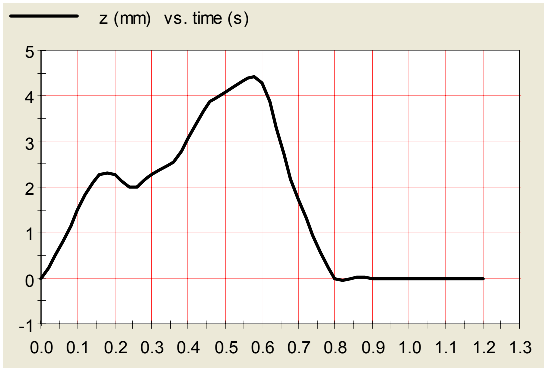
Pin 3.



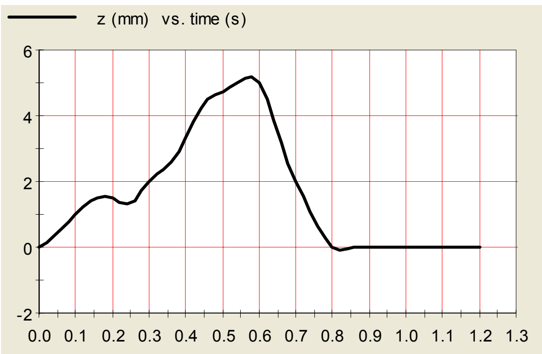
Pin 4.



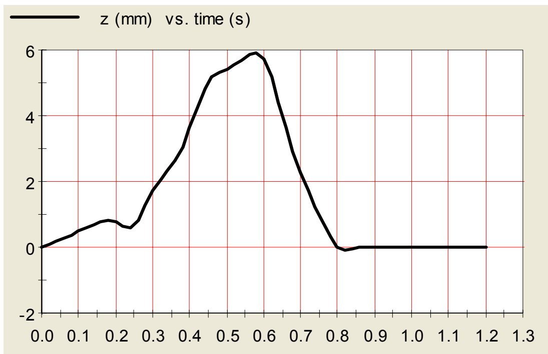
Pin 5.



Pin 6.



Pin 7.



Pin 8.

Fig. 9. Plots of pin displacement

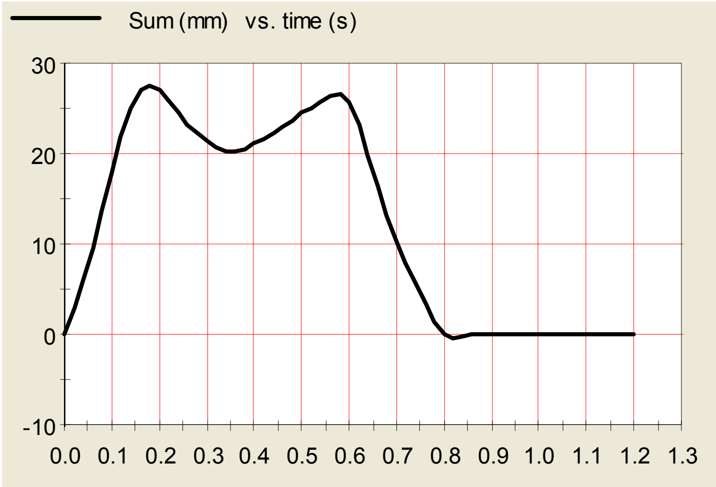


Fig.10: Sum of pin displacements.

The study performed using the computer 3D model illustrates a possibility to design and manufacture a device that would control foot loading and simulate human normal walk.

4. Conclusions

1. A 3D computer model of the mechanical module of a mechatronic system was designed. The system is used for foot receptor activation of patients - paraplegics.
2. A law of control of the two motors of a specific device was synthesized. The device is used to realize a specified foot loading during the execution of normal walk.
3. Computer simulation of the device motion was performed, adopting a synthesized law of control.
4. Displacements of the executive elements (pins) were obtained from motions of the computer model. Those displacements accomplish with sufficient accuracy the task of foot loading during walk in norm.

The conclusions of the modelling and the simulations on the parametrical biomechatronic model for definition of the proper laws and control of the drives of the system mechanical module for walk in norm of patients with spinal - brain damages support the successful realisation of the biomechatronic system.

This system can be applied in clinical practice for rehabilitation of patient - paraplegics.

Acknowledgement

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