ABSTRACT
Assistive devices such as passive or active orthoses are used in the cases of lower limb damage. For the design of an active orthosis, it is necessary to take into account the physical parameters of the patient and to calculate the characteristics of the actuator for joint articulation. It is important to decide which type of orthosis is most appropriate to use, as well as whether the use of orthosis improves or deteriorates human walking.

In this paper the developed computer models for analysis of data obtained from an autonomous adaptive system for actuation, data acquisition and control of active ankle-foot orthosis are presented. The system can operate in three modes: data collection mode, assistive mode and data monitoring mode.

Mathematical model of the orthosis together with the control unit and actuator for positioning of the foot orthotic segment and the dynamic system simulation is done in Matlab Simulink and SimMechanics.

A graphical program written in MATLAB receives and visualizes the data during walking, giving us the representation of the signals obtained from the autonomous system. For visualization of the signals measured in real time along a treadmill in laboratory conditions, a LabView virtual instrument has developed.

KEY WORDS
Gait analysis, Active ankle-foot orthoses, Control, Rehabilitation, Modeling

1. Introduction
Orthotic treatment is the most common method for foot-drop cases. Ankle foot orthoses are intended to support the ankle, correct deformities, and prevent further injuries. The idea of an actively powered orthotic device has been explored since the early 1980’s using hydraulic and pneumatic device. More recently, compressed gas and DC motors have been researched to provide active assistance to the individuals with paraplegia [1, 2]. An active ankle-foot orthosis with a force-controllable series elastic actuator (SEA) was also designed [3] capable of controlling orthotic joint stiffness and damping for plantar and dorsiflexion ankle motions.

Human movement analysis is currently a widespread and useful tool for both clinical practice and biomechanical research and provides a great deal of information regarding joint and segment kinematics and kinetics [4, 5]. Evaluation of time and distance parameters during walking is helpful in assessing abnormal gait, to quantify improvement resulting from interventions, or to predict subsequent events such as falls. Different approaches and computer methods for evaluation have been proposed. A gait laboratory based on camera, walkway with implanted sensors [6] or force-plates, and electromyography allows a complete gait analysis but require a dedicated laboratory. These techniques require the subjects to walk in a pre-defined specific path, and assume that data measured from only a few steps are representative of usual gait performance.

To avoid these limitations, autonomous system carried by the subject and allowing outdoor measurement have been developed, using new technology, such as a powerful microcontroller, miniature sensors, high capacity memory and small batteries [7, 8]. The proposed system is an autonomous adaptive device for actuation, data acquisition and control of active ankle-foot orthosis during normal level walking using the tactile sensors and the monitoring system for gait analysis. The device is used to help or rehabilitate persons with control disorders and weaknesses of ankle foot complex.

The developed computer models for personalized active orthosis design and analysis, for analysis of data obtained from the autonomous system, for estimation and correction of gait parameters during walking are described in this work.

It is important to decide which type of orthosis is most appropriate to use. For that purpose the following models are created:
- personalized model of an orthosis in Matlab SimMechanics for calculation of the torque required to drive the foot segment using anthropometric data of the human,
- personalized model of an active orthosis in Matlab Simulink together with the control unit and actuator for positioning of the foot orthotic segment and the dynamic system simulation.

The observation of changes in the human gait is an important element in the healing process. It should be done periodically. The data should be compared to see if there has been improvement or deterioration in gait or whether the use of orthosis improves or make worse...
human walking. For this purpose, a graphical program for data visualisation during walking was made:
- a graphical program written in MATLAB receives and visualizes the data during walking, giving us the representation of the signals obtained from the autonomous system;
- for visualization of the signals measured in real time along a treadmill in laboratory conditions, a LabView virtual instrument has developed.

The research work combines hardware and software design of the control device with graphical interface for representation and analysis of the data acquired during human motion. The monitoring system is used to monitor temporal parameters of each phase of the gait cycle. This system provides additional information on foot position during gait and can be used for a long period of measurement.

2. Methods

Autonomous System for Control and Monitoring (ASCAM) can operate in two modes during walking: data collection mode (Fig. 1.) and assistive mode with active ankle-foot orthosis in cases of impaired ankle-foot complex (Fig. 2.). Monitoring mode is the third mode of operation, during which the signals measured during walking are displayed in graphical user interface (GUI).

The complete autonomous system consists of four primary components - Ankle foot orthoses with actuation system, Control Module with data acquisition, sensing system, communication and friendly oriented software for interpretation of the data. The sensor system is mounted into two basic components: insole for the healthy leg or ankle-foot orthosis. The acquisition unit gathers and digitizes the information from the sensors during the walking. That data is transferred through the RS-232 lines to a graphical user interface for visualization and interpretation in monitoring mode.

The Control Module Prototype has been realised by using the microcontroller ATmega128 (Atmel Co.), the featuring analog to digital converter, USART for RS232 communication and the timer with Pulse Width Modulation (PWM) output.

2.1 Assistive mode with active ankle-foot orthosis

The active ankle-foot orthosis is a system with one degree of freedom which foot segment is connected to the shank segment by a rotational joint. A direct drive actuator is attached laterally to the AFO. Control signals are received in real time from two sensor arrays incorporated in the foot part of AFO and in the insole of the healthy leg which is the basement of the control algorithm. The microcontroller estimates forward speed and modulates swing phase flexion and extension during each gait cycle by measuring the total time TL (for the left leg) and TR (for the right leg), when the foot remains in contact with the ground, in order to achieve quite normal lower limb dynamics [9, 10].

The control algorithm is based on the biomechanical interpretation of the locomotion. Four distinct positions corresponding to the phase heel strike, stance, toe-off and swing are used within a given walking cycle. The tactile sensors and the rotary potentiometer measure an ankle joint position and send signals to the microcontroller. The microcontroller receives the diagnostic information about the system from the sensors and generates the torque command to the driver. The electro-mechanical system must actively adjust the flexion of the orthosis by actuator movement and keep this position till the heel strike appears during the swing phase, where the clearance of the toe is released. Thus, the ankle torque has to be modulated from cycle-to-cycle throughout the duration of a particular gait phase.
The **Position Control** is handled by electronics according to the output of the angular position sensor RP which is feedback element attached to the moving parts of the motor assemblies to sense the velocity and position. The sensor measures the ankle joint position in real-time. That data is used in every step of the PID control algorithm in order to maintain stability when a foot load is applied [8]. The controller reads the system state \( y \) by a rotational potentiometer, subtracts the measured angle from a desired reference \( y_0 \) to generate the error value \( e \). The error is managed in three terms - the proportional \( T_p \), the integral \( T_i \), and the derivative \( T_d \) terms which are summed to calculate the controller output \( u(t) \):

\[
u(t) = k_p e(t) + \frac{1}{T_i} \int_0^t e(\tau) d\tau + T_d \frac{de(t)}{dt}.
\]

**Control signals** from footswitches have been used to detect the gait events in real time which is the basement of the control algorithm. The footswitches consisted of two pairs of tactile sensor arrays, placed under the foot: TL1, TR1 beneath the heel and TL2, TR2 beneath the toes. For each pair, the sensors were connected in series and the outputs were recorded by Control module. During heel strike and toe-off the signals are subjected to detect changes. By these signals the exact time of heel strike and toe-off was obtained.

![Figure 3. States and transitional conditions](image)

The microcontroller collects the data from tactile sensors on four VS buffers, and creates a serial bitstream for transfer. The measured angles correspond to the rotation of 30° and obtained potentiometer signals are with sensitivity of 4 mV/deg. The signals were digitized (10 bit) at a sampling rate of 120 kHz by a microcontroller and stored on ADC buffer. At the end of the recording the data were transferred to the computer for analysis.

### 2.2 Data collection mode with metal hinge joints and two insoles

Lower limbs movement during walking was measured using signals from sensors. For measuring the ankle angle of rotation two custom made metal hinge joints were used. A potentiometer is mounted on the hinge joint, coinciding with the axis of rotation sensors. Hinged joints are attached laterally of both ankles. This mechanical motion capture system was used for measuring the patient’s ankle motion at real-time during walking (Fig. 4).

![Figure 4. a) Metal hinge joints with mounted rotational potentiometers; b) Two tactile sensor arrays are incorporated in the insole.](image)

For detecting the precise moments of heel strike (when the foot first touches the floor) and toe-off (when it takes off) during that cycle special shoes were used. Footswitches consisted of two pairs of tactile sensor arrays, placed under the foot beneath the heel and the toe have been used to detect the gait events in real time.

For visualization of the signals measured during walking along a treadmill in laboratory conditions, a LabView virtual instrument has developed. The data from sensors were collected with multifunctional (DAQ) module (NI-USB-6211, National Instruments and LabVIEW).

### 3. Mathematical model of the system.

**System analysis**

The main goal of the control system is to produce the required motor torque for the ankle actuation. But each different person has different physical parameters. Therefore a mathematical model of the system has been designed for calculation of the required motor torque for ankle actuation. The simulation of the dynamic system is done in Simulink and SimMechanics MATLAB (Fig. 5).

![Figure 5. Kinematic model of the personalized AFO in SimMechanics MATLAB](image)
This model represents the orthosis by two Body blocks connected by rotational (hinge) joint block: Body1 (shank) and Body2 (foot). The model can be personalized using the physical parameters of the patient and estimate the torque required to rotate the foot about the ankle joint assuring flexion/extension. The foot parameters are known from the conventional anthropometric tables.

To position the foot, we enforce the appropriate angle between the shank and the foot. We simulate the model in Inverse Dynamics mode to compute the joint torque required to rotate the foot in desired position. During the simulation the geometry of the orthosis is presented as a double pendulum. Once we know the computed torque, we can calculate the required dynamic motor torque and to decide which is the correct motor with appropriate parameters for joint actuation.

\[ T_z = T_d - T_c - T_g \]  \hspace{1cm} (2)

\[ T_d = (J_c + md^2)\ddot{q} + kq + mgd \sin q \]  \hspace{1cm} (3)

where \( T_d \) is the driving torque; \( T_c \) – the torque caused by the friction; \( T_g \) - torque caused by the gravity; \( J_c \) is the Body2 (foot) inertia moment; \( q \) – generalized coordinate; \( m \) - Body2 mass is sum of masses of the foot, \( m_1 \), the orthosis foot segment, \( m_2 \) and the actuator, \( m_3 \).

It is possible to develop differential equations that describe the behaviour of the DC actuator. The ultimate goal is to control the angular velocity by varying the applied voltage.

\[ \frac{di}{dt} = -\frac{R}{L}i(t) - \frac{K_b}{L}\omega(t) + \frac{1}{L}u_{app}(t) \]  \hspace{1cm} (4)

\[ \frac{d\omega}{dt} = -\frac{1}{J}K_ao(t) + \frac{1}{J}K_{ai}(t) \] \hspace{1cm} (5)

where \( K_a \) is the armature constant of the motor; \( K_b \) is the electromotive force constant; \( K_i \) is a linear approximation for viscous friction; \( J \) is the inertia of a body.

Having the differential equations we can develop a state-space representation of the DC actuator as a dynamic system in Matlab [6]. The current \( i \) and the angular velocity \( \omega \) are the two state parameters of the system. The applied voltage, \( u_{app} \), is the input to the system, and the angular velocity \( \omega \) is the output.

\[ \frac{d}{dt} \begin{bmatrix} i \\ \omega \end{bmatrix} = \begin{bmatrix} \frac{R}{L} & -\frac{K_b}{L} \\ \frac{K_n}{J} & \frac{K_f}{J} \end{bmatrix} \begin{bmatrix} i \\ \omega \end{bmatrix} + \begin{bmatrix} 1 \\ 0 \end{bmatrix} u_{app}(t), \]  \hspace{1cm} (6)

\[ y(t) = [0 \ 1] \begin{bmatrix} i \\ \omega \end{bmatrix} + [0]u_{app}(t) \]  \hspace{1cm} (7)

Giving the nominal values for parameters we can obtain the transfer function of the actuator.

Once we know the actuator parameters and computed torque, we can verify that this is the correct answer of the system simulation by analyzing driven angular motion for the articulation of the ankle joint (foot) in Matlab Simulink.

![Figure 6. Active AFO model with PID control in MATLAB Simulink](image)

The graphical representation of the active AFO model with PID control is done in MATLAB Simulink in Fig. 6. The actuator is represented by its transfer function.

4. Monitoring system for gait analysis: Experimental results

Presented autonomous system can be used for estimation of spatio-temporal parameters during long periods of walking. This method based on the biomechanical interpretation of locomotion is proposed to compute the values of temporal gait parameters and angular velocity from sensor signals. The system allows collection of data from sensors mounted under the heel and the toes part of the insole and orthosis, from two insoles or from two orthoses depending of the used mode.

4.1 Laboratory model with orthosis

The laboratory model of orthosis with hinge joint and attached laterally direct drive actuator is designed in order to test the control algorithm and system functionalities. The orthosis is with tactile sensors mounted under the heel and the toes (TR1, TR2) and restricted in the ankle joint to +/- 20 degree. A healthy subject wearing orthosis on the right leg and insole with mounted sensors on the left leg (TL1, TL2) performs different trials of slow and normal level walking.
Data Acquisition Unit. The controller collects the following parameters: ankle joint angles, tactile sensors signals and foot velocities and creates a serial bitstream for transfer. The data are collected on six buffers.

Communication and Graphical User Interface. Using USART the controller transmits data packet to the PC through the RS232 serial interface. A graphical module written in MATLAB receives the data and visualizes it in its own window, giving us the representation of the signals. The developed software allows different mathematical operations with the data, visualization and printing of the results, graphics and tables.

Gait analysis: gait temporal parameters estimation

In order to compute the temporal parameters such as the duration of swing, single and double stances during a gait cycle, it is necessary and sufficient to determine for each leg the precise moments of heel strike (when the foot first touches the floor) and toe-off (when it takes off) during that cycle. Although the duration varies according to various parameters such as subject's velocity or the presence of limping due to a painful articulation, they can always be localized. To successfully achieve this task, our system is well adapted for gait events identification.

The time diagram for the estimation of toe-off and heel-strikes for one locomotion cycle is shown in Fig.7. The gait cycle started by left toe-off (TL1) and ended by left heel strike (TL2).

\[
\begin{align*}
T &\text{ - Duration of each gait cycle (measured at right or left leg)} \\
TL &\text{ - Left stance (time between left heel strike and left toe-off)} \\
TR &\text{ - Right stance (time between right heel strike and right toe-off)} \\
T_{d1} &\text{ - Initial double support (time between right heel strike and left toe-off), known also as left double thrust support time} \\
T_{d2} &\text{ - Terminal double support (time between left heel strike and right toe-off), known also as right double thrust support time} \\
T_d &\text{ = } T_{d1} + T_{d2} \text{ - Double support}
\end{align*}
\]

Every temporal parameter of a gait cycle can be easily computed as percentage of gait cycle.

Gait model: spatial parameters estimation

We propose a double segment gait model involving both shank and foot. In this model, the swing phase is considered as a double pendulum model, while the stance phase is considered as an inverse double pendulum model. Rotary potentiometer provides ankle angle signal ADC (in millivolts) during swing and stance phase. Based on a mechanical model, the medio-lateral rotation of the lower limbs during stance and swing, the stride length and velocity are estimated by integration of the angular velocity.

4.2 Experimental results with metal hinge joints and two insoles

Measurements were taken from healthy subject. Each trial includes, at least 20 gait cycles performed with normal level walking. The sensors work together to detect walking over one given interval of time and to collect the following parameters: ankle joint angles (Fig.8.a) and foot contacts (heel and toe) (Fig.8.b).

(a) Ankle angle rotation (potentiometer data in volts)

(b) Tactile sensors results

Figure 8. Visualization of human motion data (in LabView)

The signals are shown and recorded in LabView virtual instrument. Signals line0 and line1 are recorded digital signals from the switches mounted under the heel and the toes part of the left leg insole while signals from the right leg are line2 and line3. The first one transition of line0 signal from 1 to 0 shows the heel strike component (left stance) and the second one transition of line1 signal from 0 to 1 shows the toe-off component (for the left leg). The values of the signals were detected in milliseconds.

The graphic on Fig.9 shows the kinematics of the
ankle. Ankle angle rotation (potentiometer data in volts) is shown with the presence of peaks during the heel strike flexion or negative peaks during toe-off and extension. The range of the measured angles correspond to the rotation of 30° and obtained potentiometer signals are with sensitivity of 4 mV/deg. It is obvious that the algorithm will be applied in the same way for both legs.

Figure 9. Ankle angle rotation (potentiometer data in volts). The peaks show the heel strike flexion and the negative peaks show toe-off and extension.

Gait analysis: gait temporal parameters estimation

The correspondence between temporal events detected by sensors is shown for several consecutive gait cycles. Left and right heel strikes (respectively toe-off) were detected from tactile sensors. The temporal parameters (over the N cycles) of left and right gait cycle time (T), left and right stance (SL and SR), initial and terminal double stance (Td1, Td2) were computed for each trial.

The signals can be processed and visualised in different ways. They also can be displayed in Matlab or Excel charts and tables.

In Fig.10 are shown the following temporal parameters obtained by mathematical comparison of the signals in LabView virtual instrument:

- Left stance, SL - time between left heel strike and left toe-off obtained by mathematical comparison of the signals from tactile sensors TL0 and TL1,
- Right stance, SR - time between right heel strike and right toe-off obtained by mathematical comparison of the signals from tactile sensors TR0 and TR1.

In Fig.11 are shown the Initial and Terminal double supports (Td1, Td2) obtained by mathematical comparison of the temporal parameters SL and SR (Left stance and Right stance).

Figure 11. Calculation of the Initial and Terminal double supports (Td1, Td2) temporal parameters

The results from two different series of samples during normal walking assume that a gait cycle is always less than 2 s. The toe-off detection was found to be in the intervals of 0.75 s. Finally, at each cycle, the actual heel strike, HS and toe-off, TO must be so that their difference is considered reasonable for walking verifying the condition 0.1 s < HS – TO < 2.5 s. As it is illustrated, the estimated velocity is almost constant 1.11 m/s (4 km/h).

The change of gait cycle provides a corresponding change of stride length in order to maintain the velocity constant.

The observation and analysis of changes in the human gait should be done periodically. The data should be compared to see if there has been improvement or deterioration in gait or whether the use of orthosis improves or make worse human walking.

5. Discussion

Presented autonomous system for control and monitoring is well adapted for gait events identification. The temporal parameters were estimated based on the biomechanical interpretation of locomotion and assuming that in straight walking, right and left stride length are not equal. Gait events can be explained by looking at the patterns of the tactile sensor signals and ankle angle rotation of the both foot. Most of the subjects had slightly asymmetric stride length and velocity therefore the right and left ankle rotation can be estimated separately by using a hinge joint with mounted potentiometer. In this way, foot contacts and ankle rotation angle give insight into the recognition of the periods of ankle flexion and extension during gait cycle. These techniques provide satisfactory results for
normal walking as well as assessing subjects with abnormal gait.

The presented device for control of active ankle-foot orthosis integrates biomechanics based algorithms with active control system. The autonomy of the developed system has been demonstrated presenting experimental data during walking. The system controls the orthosis functionalities, records the data received from sensors during the gait and transfers recorded data to graphical user interface for visualization and future analysis.

The controller is battery powered and can collect data up to 8 Mbytes. Additional memory or lower sampling rate can be used to increase the period of recording. The system is open - we can add more inputs for signals measured articulation of the knee or upper limb.

There is a problem with the attachment of hinge joints. They are rigid and movable during walking. Therefore, they are covered with foam and fastened by strips.

6. Conclusion

The developed device for control of AAFO provides broad information for both control and gait analysis. The data from the sensors are used in every step from the control algorithm. The actuator joint torque is automatically modulated in order to optimize the heel-to-forefoot transition during the stance or the swing phase of walking. The experimental data discussed in this paper can be used in cases of the drop foot treatment and lower limb rehabilitation to enhance the AAFO functional performance and to improve the patient gait.

Based on the above, we can say that the proposed computer models for gait identification and analysis of data obtained from the autonomous system appears a promising monitoring tool for several purposes. First, it allows measurements of gait features during a long period of walking and thus supplies the stride-to-stride variability of gait. In addition this system can be used for orthosis actuation therefore provides information that is more likely to reflect the actual performance of the subjects.

It can be used in many clinical applications such as outcome evaluation after knee and hip replacement, or external prosthesis adjustment for amputees. In elderly subjects, this system can also be proposed as a diagnostic tool for abnormal gait analysis, as a predictor tool for fall risk estimation, or as a monitoring tool to assess progress through rehabilitation.

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