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# DEVICE FOR CONTROL OF ACTIVE ANKLE-FOOT ORTHOSIS AND MONITORING SYSTEM FOR GAIT ANALYSIS*

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Abstract. The aim of this paper is to present an intelligent device for control of an active ankle-foot orthosis, which can be used for assisting and rehabilitation in the cases of an injured ankle-foot complex together with a graphical interface for representation and analysis of the data acquired during human motion
The device proposed for control of the Ankle Foot Orthosis (AFO) consists of a microcontroller, an actuator and a sensor system. The microcontroller estimates forward speed and modulates swing phase flexion and extension during the gait in order to assure automatic adaptation of the AFO joint torque by using position control with feedback. Data acquired from the sensors is transferred to a graphical user interface for visualization, interpretation and analysis in monitoring mode.
Key words: control, active ankle-foot orthoses, biomechanics, rehabilitation.

## 1. Introduction

The ankle foot orthoses are intended to support the ankle, correct deformities, and prevent further injuries. Orthotic treatment is the most common method for the foot-drop cases. The idea of an actively powered orthotic device has been explored since the early 1980's by using a hydraulic and pneumatic device. More recently, a compressed gas and DC motors have been researched

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to provide an active assistance to individuals with paraplegia [1, 2]. The active ankle-foot orthosis with a force-controllable Series Elastic Actuator (SEA) is also designed [3] capable of controlling orthotic joint stiffness and damping for plantar and dorsiflexion ankle motions.

In this paper we propose an autonomous adaptive device for actuation, data acquisition and control of an 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 the ankle foot complex.

## 2. Methods

The complete autonomous system consists of four primary components – sensing, data acquisition, communication and friendly oriented software for interpretation of the data (Fig. 1).

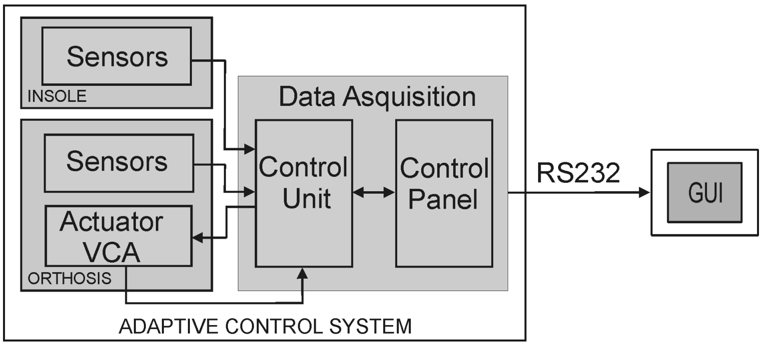



Fig. 1. Autonomous control and monitoring system with active ankle-foot orthoses

The sensor system is mounted into two basic components: insole for the healthy leg and ankle-foot orthoses. 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 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. A Proportional-Integral-Derivative (PID) control with position feedback is used to estimate the trajectory of the foot and positioning the actuated foot segment of AFO when the foot rotates about the ankle. The microcontroller estimates forward speed and modulates swing phase flexion and extension during each gait cycle in order to assure automatic adaptation of the joint torque [4, 5].

Four distinct positions corresponding to the phases *heel strike, stance, toe-off and swing* are used within a given walking cycle.

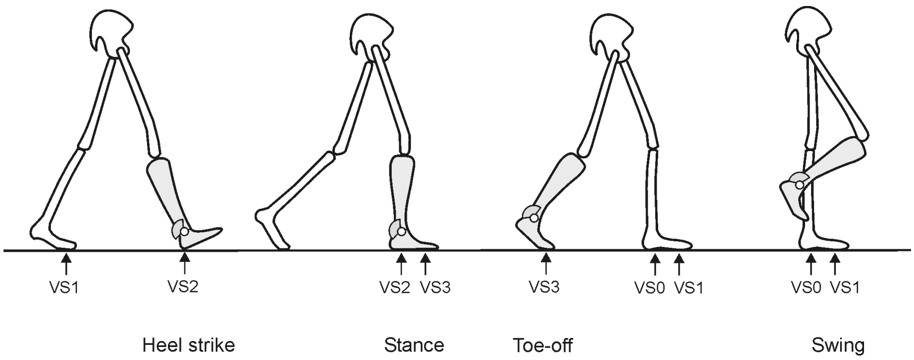



Fig. 2. States and transitional conditions

The electro-mechanical system must actively adjust the flexion of the orthosis by actuator movement during the swing phase, where the clearance of the toe is released and keep this position till the heel strike appears. Thus, the ankle torque has to be modulated from cycle-to-cycle throughout the duration of a particular gait phase.

### 3. Control module prototype

The active ankle-foot orthosis is an electro-mechanical system controlled by a control module. The control module prototype (Fig. 3) has been realised by using the microcontroller ATmega128 (*Atmel Co.*), that has analog to digital converter, USART for RS232 communication and the timer with Pulse Width Modulation (PWM) output [4, 5]. The PWM channel is connected to the driver to control the direction and speed of the motor by varying the duty cycle of the PWM output. The speed and the torque of the motor can be varied by varying the current flow thought the coil.

The *Control algorithm* is based on the biomechanical interpretation of the locomotion. 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.

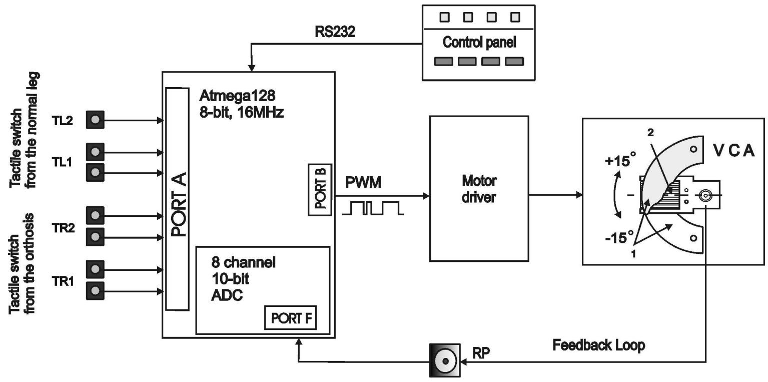



Fig. 3. ATmega128 controlling a VCA

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 (Fig. 4).

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

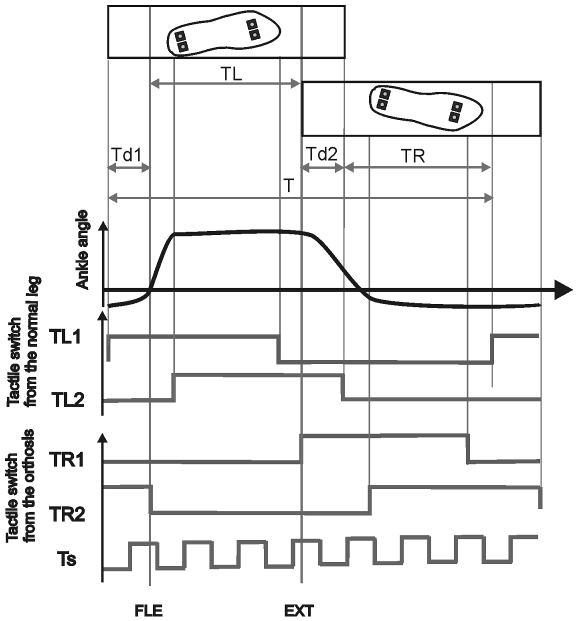



Fig. 4. Signals from the external sensors

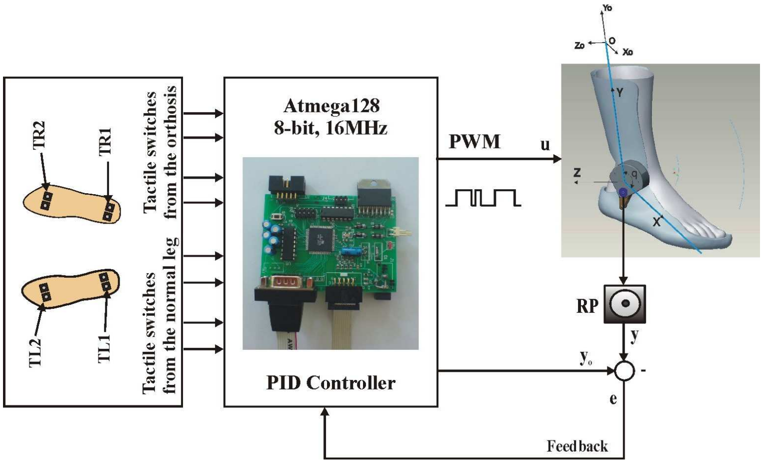



Fig. 5. Autonomous Control System with Active Ankle-Foot Orthoses

sensor measures the ankle joint position in real-time (Fig. 5). That data is used in every step of the PID control algorithm in order to maintain stability when a foot load is applied [7]. 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 will be managed in three terms – the proportional $T_p$, the integral $T_i$, and the derivative $T_d$, terms are summed to calculate the output based on the PID algorithm. We obtain the final form of the PID algorithm defining $u(t)$ as the controller output:

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

We obtain the discrete form of the controller approximating the integral and the derivative terms:

$$u(n) = K_p e(n) + K_i \sum_{k=0}^{n} e(k) + K_d (y(n) - y(n-1)), \tag{2}$$

where $n$ is the discrete step at time $t$.

## 4. Monitoring system for gait analysis

*Data Acquisition Unit.* The controller collects the following parameters: ankle joint angles, tactile sensors signals and foot velocities and creates a serial

Table 1. Buffers for data collection

| Signal | PWM | ADC | TL1 | TL2 | TR1 | TR2 |
|---|---|---|---|---|---|---|
| Bufer | PWM_buf | ADC_buf | VS0_buf | VS1_buf | VS2_buf | VS3_buf |
| File | *sigPWM* | *sigADC* | *sigVS0* | *sigVS1* | *sigVS2* | *sigVS3* |

bitstream for transfer. The data is collected on four VS buffers, ADC buffer, PWM duty-cycle buffer:

*Communication and Graphical User Interface.* The controller transmits data packet to the PC through the RS232 serial interface using USART. The graphical module written in MATLAB receives the data and visualizes it in its own window giving us the representation of the signals (Fig. 9). The software developed allows different mathematical operations with the data, visualization and printing of the results, graphics and tables.

## 5. System analysis model

The main goal of the control system is to produce the motor torque required for the ankle actuation. But each different person has different physical parameters. Therefore, the mathematical model of the system has been designed for calculation of the required motor torque for the ankle actuation [6]. The simulation of the dynamic system is done in Simulink and SimMechanics MATLAB. This model represents the orthosis by two Body blocks connected by rotational (hinge) joint block: Body1 (shank) and Body2 (foot) (Fig. 6(a)).

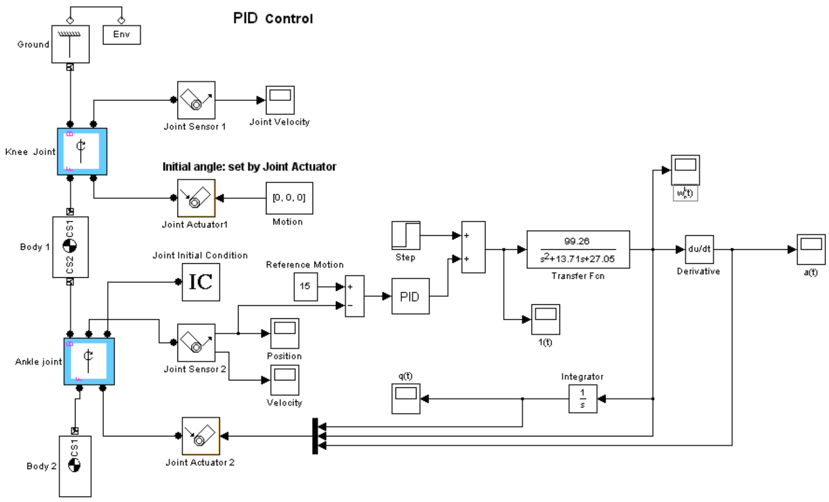
The model can be personalized by using the physical parameters of the patient and the torque required to rotate the foot about the ankle joint assuring flexion/extension will be calculated. The foot parameters are known from the conventional anthropometric tables.

We enforce the appropriate angle between the shank and the foot in order to position the foot. We simulate the model in *Inverse Dynamics mode* to compute the joint torque required to rotate the foot in desired position. The geometry of the orthosis is presented as a double pendulum during the simulation (Fig. 6(b)). We can calculate the required dynamic motor torque for the computed torque in order to choose the correct motor with appropriate parameters for joint actuation:

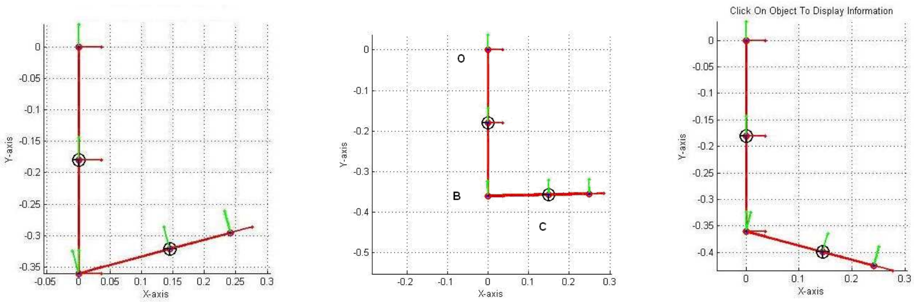
$$T_z = T_d - T_c - T_g, \tag{3}$$

$$T_d = (J_c + md^2)\ddot{q} + k\dot{q} + mgd \sin q, \tag{4}$$

where $T_d$ is the driving torque; $T_c$ is the torque caused by the friction; $T_g$ is the torque caused by the gravity; $J_c$ is the foot (Body2) inertia moment; $q$ is




(a)




(b)

Fig. 6. (a) Electromechanical system model with PID control in MATLAB; (b) The geometry of the orthosis is presented as a double pendulum during the simulation in MATLAB

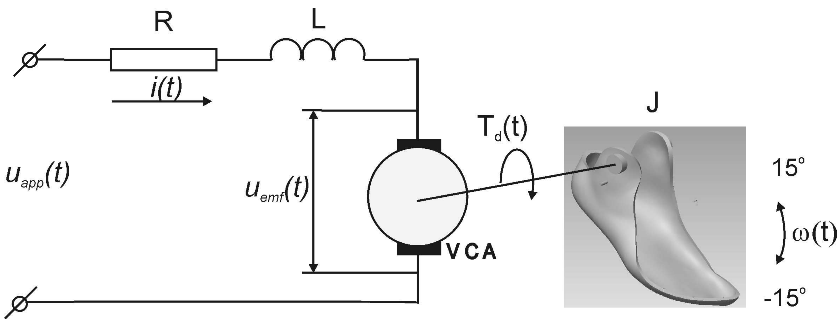



Fig. 7. A simple model of a DC actuator driving an inertial load

the generalized coordinate; $m$ is the Body2 mass, sum of masses of the foot $m_1$, the orthosis foot segment $m_2$, and the actuator $m_3$.

The model of direct drive DC actuator (Fig. 7) driving an inertial load (foot) shows the angular velocity of the foot $w(t)$, as the output, and applied voltage $u_{app}(t)$, as the input. The resistance of the circuit is $R$, and the self-inductance of the armature is $L$.

The differential equations that describe the behaviour of the DC actuator are developed by using this model. The ultimate goal is to control the angular velocity by varying the applied voltage:

$$\frac{di}{dt} = -\frac{R}{L}i(t) - \frac{K_b}{L}\omega(t) + \frac{1}{L}u_{app}(t), \tag{5}$$

$$\frac{d\omega}{dt} = -\frac{1}{J}K_f\omega(t) + \frac{1}{J}K_m i(t), \quad \omega(t) = \frac{dq}{dt}, \tag{6}$$

where $K_m$ is the armature constant of the motor; $K_b$ is the electromotive force constant; $K_f$ is a linear approximation for viscous friction; $J$ is the inertia of a body.

We can develop a *state-space* representation of the DC actuator as a dynamic system in Matlab [6] using the differential equations. 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:

$$\frac{di}{dt}\begin{bmatrix} i \\ \omega \end{bmatrix} = \begin{bmatrix} -\frac{R}{L} & \frac{K_b}{L} \\ \frac{K_m}{J} & \frac{K_f}{J} \end{bmatrix} . \begin{bmatrix} i \\ \omega \end{bmatrix} + \begin{bmatrix} \frac{1}{L} \\ 0 \end{bmatrix} .u_{app}(t), \tag{7}$$

$$y(t) = \begin{bmatrix} 0 & 1 \end{bmatrix} . \begin{bmatrix} i \\ \omega \end{bmatrix} + [0]\,.u_{app}(t). \tag{8}$$

We can obtain the transfer function of the actuator giving the nominal values for parameters. The actuator is represented by its transfer function in Fig. 6(a).

Thus, we know the actuator parameters and computed torque, and 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*.

**6. Experimental results**

The control module proposed is designed and tested. 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 (Fig. 8). The orthosis is restricted in the ankle joint to ±20 degree.

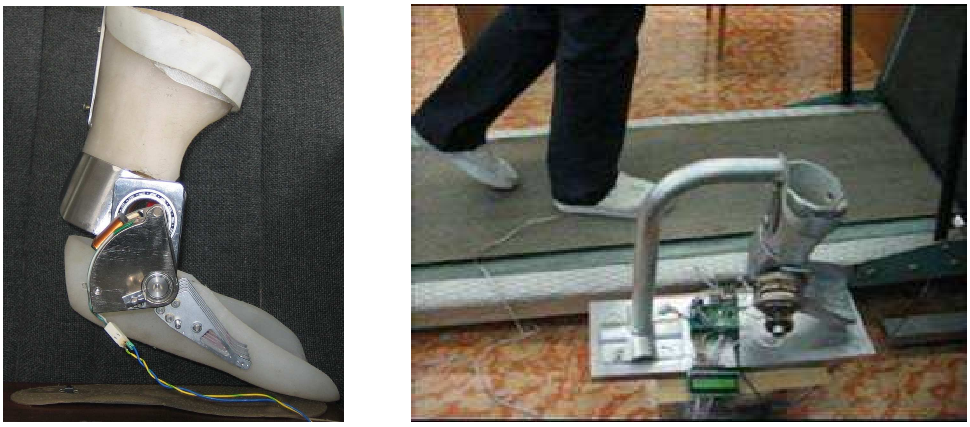


Fig. 8. AAFO with the hinge joint and attached laterally direct drive actuator

A healthy subject equipped with the sensors mounted under the heel and the toes part of the insole (TR1, TR2 for the right leg and TL1, TL2 for the left leg) performs different trials of slow and normal level walking. The motion of the orthosis is observed during walking – if the time and phase parameters of the orthosis coincide with those of the right leg.

The sensors work together to detect walking over one given interval of time and to collect the following parameters: ankle joint angles, foot (heel and toe) contacts and foot velocities. The microcontroller collects sensor data in four VS buffers, ADC buffer and PWM duty-cycle buffer.

The data is transmitted to the PC through the RS232 serial interface in monitoring mode. The graphical program module receives and visualizes the data giving us the representation of the signals (Fig. 9). A rotary potentiometer provides an ankle angle signal ADC (in millivolts) during swing and stance phase. Signals VS0 and VS1 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 VS2 and VS3.

The block diagram model in *Matlab Simulink* is a graphical representation of the mathematical model of the dynamic system. The orthosis is represented by two Body blocks and one Revolute Joint block, and the actua-

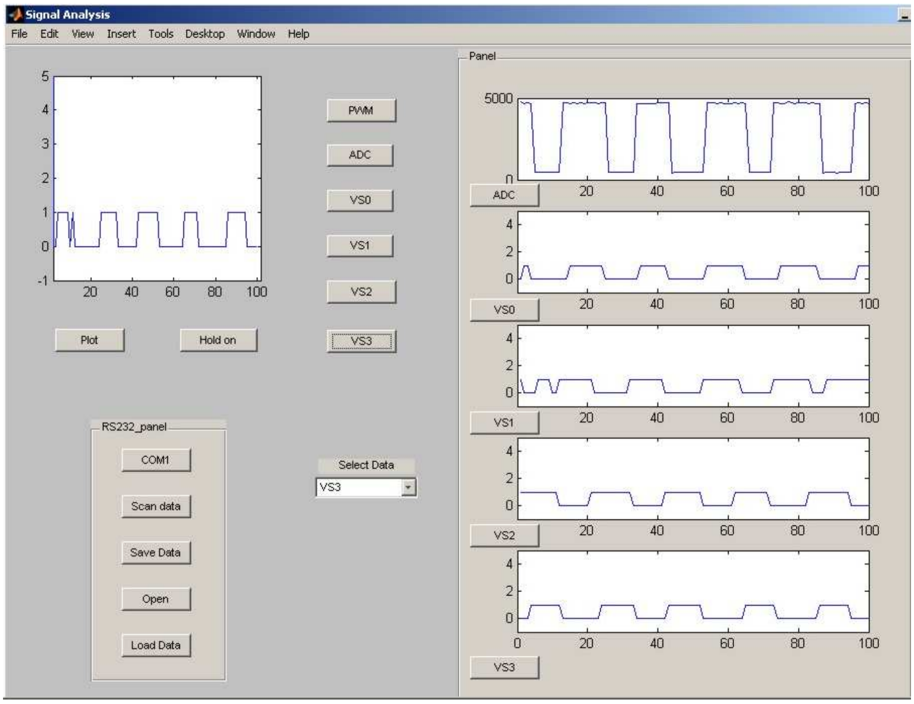



Fig. 9. Graphical program written in MATLAB for visualization of human motion data

tor is represented by its transfer function. The model is personalized by using the physical parameters of the patient and the torque required to rotate the foot about the ankle is estimated:

$m = m_1 + m_2 + m_3 = 1$ kg is the Body2 mass (sum of the foot mass, orthosis mass and actuator mass);

$J = J_c + md^2 = 3.7 + 0.7 \times 225 = 0.01612$ (kg.m$^2$) is the Body2 Inertia about ankle axis;

$d = 0.15$ m is the distance between the ankle joint and Body2 centre of mass;

$J_c = 3.7$ kg cm$^2$ is the inertia moment, about an axis through Body2 centre of mass, parallel to the ankle axis.

The calculated by the system driving torque is $T_d$ =1.4214 Nm. Thus, we know the computed torque, and we can choose the correct motor with appropriate parameters for joint actuation. We obtain the transfer function for the given nominal values of the parameters of the selected Rotary Voice

Coil Actuator $W(s)$:

$$W(s) = \frac{99.26}{s^2 + 13.71s + 27.05}.$$

We verify that this is the correct answer of the system simulation analyzing angular motion of the ankle joint in *Matlab Simulink*.

## 7. Discussion

The device presented for control of the active ankle-foot orthosis integrates algorithms based on the biomechanics with an active control system. The autonomy of the developed system has been demonstrated by 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 device developed for control of AAFO provides broad information for both the control and the gait analyses. The data from the sensors is 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 the cases of the drop foot treatment and the lower limb rehabilitation to enhance the AAFO functional performance and to improve the patient gait.

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