

GENERAL MECHANICS

PROPULSION SYSTEM WITH PNEUMATIC ARTIFICIAL MUSCLES FOR POWERING ANKLE-FOOT ORTHOSIS*

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ABSTRACT. The aim of this paper is to present the design of device for control of new propulsion system with pneumatic artificial muscles. The propulsion system can be used for ankle joint articulation, for assisting and rehabilitation in cases of injured ankle-foot complex, stroke patients or elderly with functional weakness.

Proposed device for control is composed by microcontroller, generator for muscles contractions and sensor system. The microcontroller receives the control signals from sensors and modulates ankle joint flexion and extension during human motion. The local joint control with a PID (Proportional-Integral Derivative) position feedback directly calculates desired pressure levels and dictates the necessary contractions.

The main goal is to achieve an adaptation of the system and provide the necessary joint torque using position control with feedback.

KEY WORDS: Control, artificial muscles, joint articulation, rehabilitation robotics.

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1. Introduction

Nowadays, robots are widely used in the immediate surroundings of people and there is a need for safety against all possible accidents. Therefore, soft actuators are being developed. Pneumatic systems are an interesting alternative for the actuation of legged robots and gait therapy devices. Such soft actuators offer a natural compliance making them human friendly.

Several concepts of pneumatic artificial muscles (PAM) have been developed over time, some examples are Romac muscle [1], Baldwin muscle [2], pleated pneumatic artificial muscle [3] and the best known type is the so called McKibben muscle [4]. This muscle was introduced by McKibben for orthotic applications in the fifties. Several forms of this type of muscle have actually been commercialized by different companies such as Bridgestone Co. [5], the Shadow Robot Company [6], Merlin Systems Cooperation [7] and Festo [8]. The interest in these actuators is growing increasingly and several groups all over the world use McKibben muscles in various robotic and medical applications [9–12].

Actuator technologies that are used in gait therapy devices such as pneumatic and hydraulic actuators can provide the required power with a small device but are impractical as portable devices, because they require separate pumps or other air supply [13, 14]. The question is, whether it is possible to design small pumps which will be filled with air during walking. These natural pumps could be placed under the soles of both feet and will generate compressed air under the influence of the weight of the human body. This compressed air will be used to power the pneumatic muscles and to generate muscle contractions. As energy consumption is an important issue, it remains important to use the natural dynamics of human movement for energy storage in the form of compressed air. It is known that a 75 kg subject produces, at the ankle, on average 26J of energy during one stride at a normal cadence and stores and releases about 9J of energy [15] during stance.

The goal of this work is to develop an active lower limb orthosis which exploit the natural dynamics of human movement by trying to harvest the energy generated during walking and use it as air supply for pneumatic muscles. This orthosis should be able to adapt the natural dynamics as a function of the imposed walking motion. It is believed that these pneumatic actuators have characteristics, which can be well adapted, controlled and used for assisting of human planar walking.

In this context, the development of full orthosis for lower limb with

lightweight materials and joints actuated with pneumatic artificial muscles is started. In this paper, we presented the design of ankle-foot orthosis (AFO) with pneumatically actuated hinge joint, and self supplied pumps with compressed air.

2. Methods

The goal of this work is to create a lightweight active orthosis, which is able to exploiting the adaptable passive behaviour of the pneumatic artificial muscles in order to maintain adequate stiffness in the joints and reduce energy consumption using the natural dynamics of human movement for energy storage in the form of compressed air. Stiffness can easily be changed by changing the applied pressures [16–17] through implementation of McKibben’s type of pneumatic artificial muscles. The airmass consumption and hence the energy consumption can be minimized by calculating an optimal stiffness depending on the desired trajectory. The natural unforced motion of the system is then very close to the desired motion [18].

A pneumatic artificial muscle is essentially a volume, enclosed by a reinforced membrane, that expands radially and contracts axially when inflated with pressurized air. Hereby, the muscle generates a unidirectional pulling force along the longitudinal axis. When neglecting the membrane’s material deformation and the low inertial muscle properties, the generated force F is expressed:

$$(1) \quad F = -p \frac{dV}{dl},$$

where p is the gauge pressure inside the muscle, dV enclosed muscle volume changes and dl actuator length changes. The volume of the actuator increases with decreasing length until a maximum volume is reached. In Fig. 1 the concept of the McKibben muscle is given. It contains a rubber inner tube which will expand when inflated, while a braided sleeving transfers tension. Inherent to this design are dry friction between the netting and the inner tube and deformation of the rubber tube. These problems are avoided in the Festo muscles.

Depending on the geometry and type of the membrane, the specific force characteristic alters. At maximum contraction, forces become zero, and at low contraction these forces can be very high.

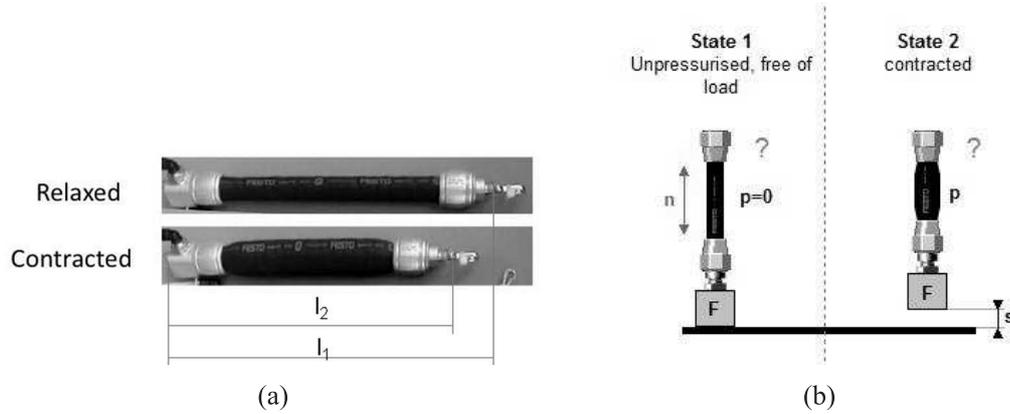


Fig. 1. The volume of the contracted actuator increases with decreasing length: (a) McKibben muscle; (b) Festo muscle

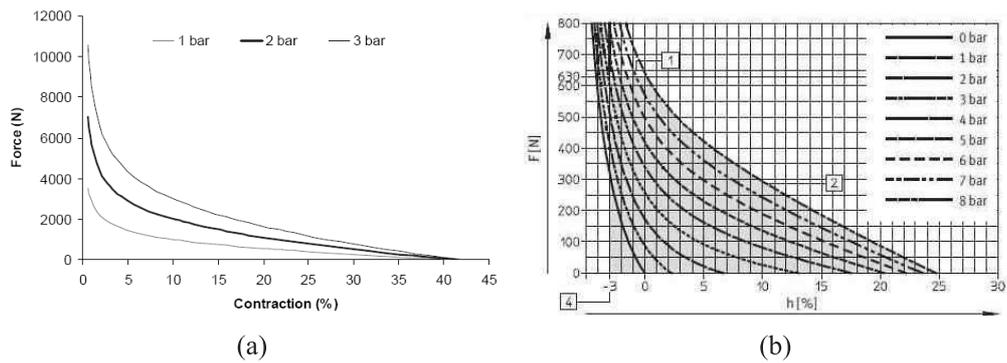


Fig. 2. Forces at pressure levels 1, 2 , 3 bar as a function of contraction: (a) PPAM diagrams, (b) Festo manufacturer diagrams

Figure 2 gives the working principle of a PAM at constant pressure [8, 17]. The graph shows the nonlinear character of the generated muscle force. For small contractions, the forces are extremely high, while for large contractions, the forces drop to zero. For the practical application, contractions will be bounded somewhere between 5 and 35% for PPAM and 0 and 25% for Festo muscles.

Typical working pressure values of Festo muscles range from 1 to 5 bar and more. Due to a threshold of pressure which depends on the rubber characteristics, these muscles do not function properly at low pressures.

Many actuator materials and devices have been put forth as “artificial muscles.” Pneumatics and hydraulics (including soft-bodied actuators such as the “McKibben Muscle” can imitate much of the performance of natural muscle and have a shape and feel similar to natural muscle, but they are noisy, difficult to control, and require a separate pump to provide the fluid energy.

In this paper, we propose an adaptive device for ankle joint actuation with custom made pneumatic muscles. The device for actuation, data acquisition and control of active ankle-foot orthosis, recently proposed by Veneva [21], was used as a basis. We use pneumatic membranes for registering foot contacts during normal level walking, for supplying with compressed air and for joint actuation.

The purpose was to confirm our new dynamic control scheme and to confirm that by harnessing the stored energy in the pneumatic membranes, motor and energy requirements were significantly reduced. Thus, we designed a pneumatically powered, controlled ankle-foot orthosis as a tool for rehabilitation and studying human locomotor adaptation. Future work will extend the concept to a hip and knee orthosis to provide assistance at other joints.

3. Description of the system

Ankle-foot orthosis is a system with one degree of freedom which foot segment is connected to the shank segment by a rotational joint. Two identical artificial pneumatic muscles are attached laterally to the ankle. The pneumatic muscles are lightweight custom made and can produce high power outputs. They are made from latex tubing surrounded by a braided polyester shell. Inflating the tubing causes the shell to expand radially and shorten axially. The tubing is positioned into two end fittings which close the muscle and provide tubing to inflate and deflate the enclosed volume. Due to its specific design, the PAM can easily work at pressures as low as 20 mbar. Muscle contraction can be more than 40 %, depending on its original dimensions. The muscle prototype has a weight of about 100 gr while it can generate forces up to 1 kN.

Two pressure membranes with a volume of 40 ml are incorporated under the foot for generation of muscle contractions under the weight during the particular gait phase.

The muscle is shown in Fig. 3 in its inflated and deflated state. A several tests were performed, at which a muscle moves up and down a load of 1 kg by a slow varying gauge pressure between 1 and 3 bars.

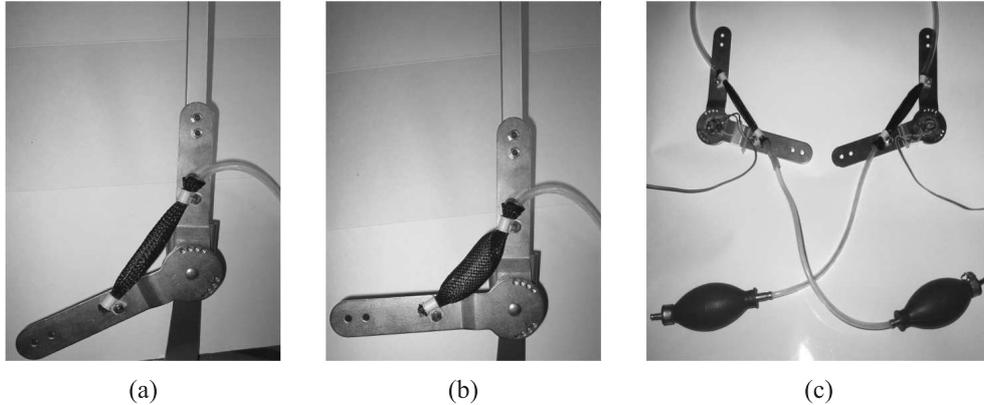


Fig. 3. Inflated and deflated state of pneumatic artificial muscle. Muscle contractions are generated by pushing pressure membranes incorporated under the foot.

The *Control algorithm* is based on the biomechanical interpretation of the locomotion [21, 22]. Within a given walking cycle, four distinct positions were used corresponding to the phases: *heel strike*, *stance*, *toe-off* and *swing*. During the swing phase, where the clearance of the toe is released, the system must actively adjust the flexion of the orthosis 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. This algorithm works well and will be used in the new system with pneumatic artificial muscles.

Four pressure membranes and tactile sensors (TR1, TR2 and TL1, TL2) are incorporated under the heel and the toes in the shoes of both legs. We push air to the muscles during each gait cycle by pressuring the membranes under the influence of weight. Inflating the tubing causes the shell to expand radially and shorten axially thus, generating muscle contractions. During each gait cycle, by pressing the corresponding tactile sensors the electrical valves are open and the air comes out decreasing the tubing, thus, generating muscle relaxation. Thereby, the ankle torque for each leg is modulated by pushing the pressure membranes of the opposite leg during the particular gait phase.

In this way, the required flexion of the ankle joint is realized by pneumatic muscles contraction using pneumatic membranes for registering foot contacts and for joint actuation. Thus, we propose an adaptive actuation of the joint harnessing the stored energy in the pneumatic element.

The control module is based on an adaptive device for actuation, data acquisition and control of active ankle-foot orthosis, recently proposed by Ve-

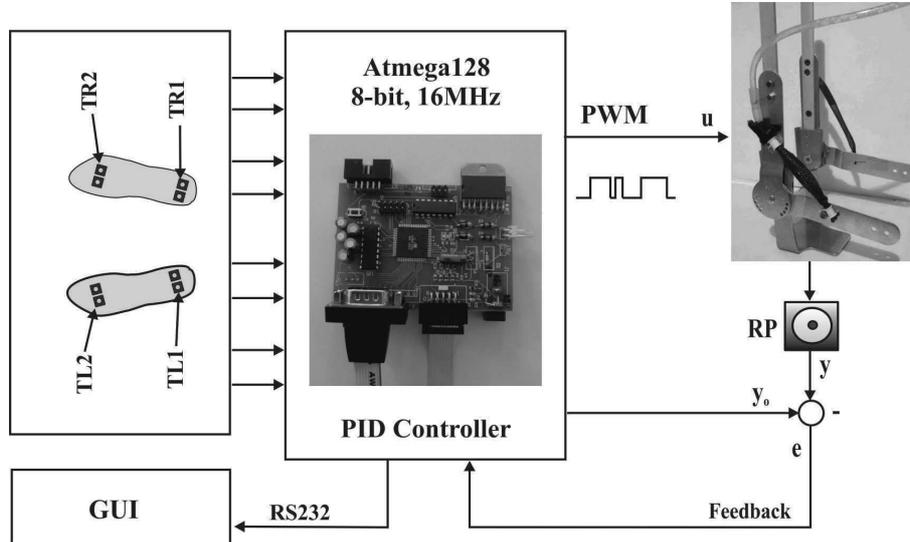


Fig. 4. Control system with ankle-foot orthoses and pneumatic artificial muscles

neva [21]. It has been realised using microcontroller ATmega128 (*Atmel Co.*). The microcontroller receives the diagnostic information about the system from the sensors and generates the signals to the valves for opening or closing. The contracted pneumatic artificial muscle generates a pulling force along the longitudinal axis. The muscle relaxation is controlled by a fast switching on/off valve. The PWM channel is connected to the driver to control the speed of the valve solenoid actuator by varying the duty cycle of the PWM output. Control signals are received in real time from sensors. The tactile sensors and a rotary potentiometer measure ankle joint position and send signals to the microcontroller. A Proportional-Integral-Derivative control with feedback was used to estimate the trajectory of the foot and positioning the actuated foot segment of AFO when the foot rotates about the ankle. During each gait cycle a microcontroller estimates forward speed and modulates swing phase flexion and extension in order to assure automatic adaptation of the joint torque.

4. Ankle joint setup

The leg consists of three parts (Fig. 5a): lower leg, upper leg and foot. The length of the i -th link is l_i , its mass is m_i and the moment of inertia about its centre of mass G_i is I_i . While the foot is in contact with the ground, this

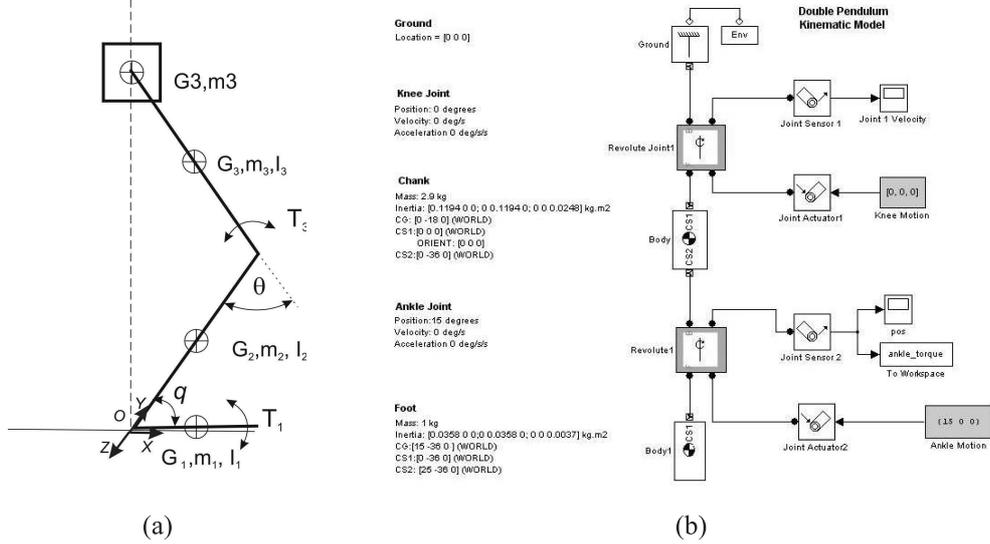


Fig. 5. Kinematics model – (a) Lower limb; (b) Personalized AFO in SimMechanics MATLAB

model has one degree of freedom (DOF) which is represented by the relative knee angle θ .

Different mechanisms will be used for the hip, knee and ankle joint actuation. Here we are going to discuss how to personalize the model of the ankle joint actuation using physical parameters of the patient, estimate the length of the muscle and point of attachment, the pulling force (respectively torque) required to rotate the foot about the ankle in appropriate angle.

The dynamic model of the system is in the swing phase and it assumes that the shank is inertially fixed (Fig. 5b). Ankle-foot orthosis is built of two segments – shank and foot. The foot segment is connected to the shank segment by a rotational (hinge) joint with a single rotational degree of freedom, which is represented by the ankle angle q . To determine the kinematics expressions of the joint system, an orthogonal XY-coordinate system is defined. The X-axis is aligned with the floor, while the vertical Y-axis is attached to the shank and the axis Z represents the ankle joint axis. To position the foot, we enforce the appropriate angle between the shank and the foot. We simulate the model in *Inverse Dynamics mode* in SimMechanics MATLAB 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 [22]. The joint

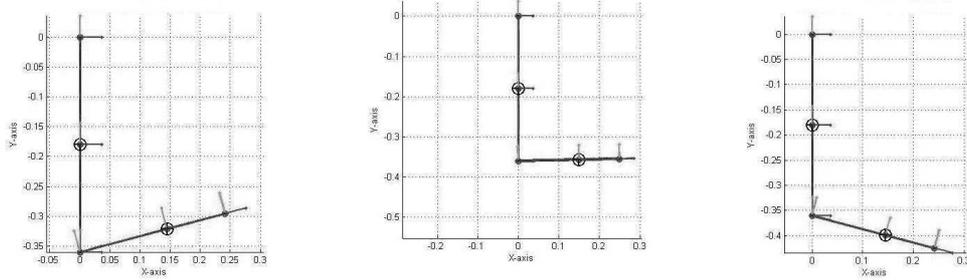


Fig. 6. Representation of the orthosis geometry as a double pendulum in SimMechanics

torque is given by following expression:

$$(2) \quad T = T_d - T_c - T_g,$$

$$(3) \quad T_d = (J_c + md^2)\ddot{q} + k\dot{q} + mgd \sin q,$$

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 foot body inertia moment; q – generalized coordinate; m – sum of masses of the foot and orthosis foot segment.

The essential parameters to be determined during the design process of the joint (Fig. 7) are the following:

- l_0 – the length of muscle when relaxed;
- l_1 – the length of muscle when contracted;
- r_1 – the distance between the origin O and the point A – the muscle attachment of the foot segment;
- r_2 – the distance between the origin O and the point B – the muscle attachment of the shank segment;
- q – the ankle angle between the vector OA and OB, with pivot point O (counter-clockwise is positive);

The contracted pneumatic artificial muscle generates a pulling force along the longitudinal axis.

$$(4) \quad \tau = r_1 F \sin q,$$

$$(5) \quad q = f(l, p).$$

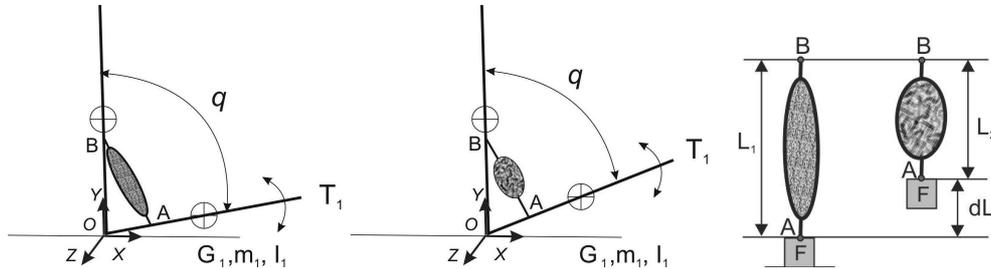


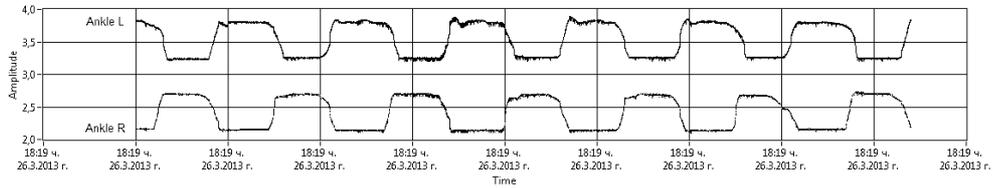
Fig. 7. Ankle joint with pneumatic muscle generated pulling force-relaxed (left), contracted (right)

The model can be personalized using the physical parameters of the patient and estimate the length of muscle and point of attachment A and B, the pulling force required to rotate the foot about the ankle and the minimum and maximum angle of joint rotation. The foot parameters are known from the conventional anthropometric tables.

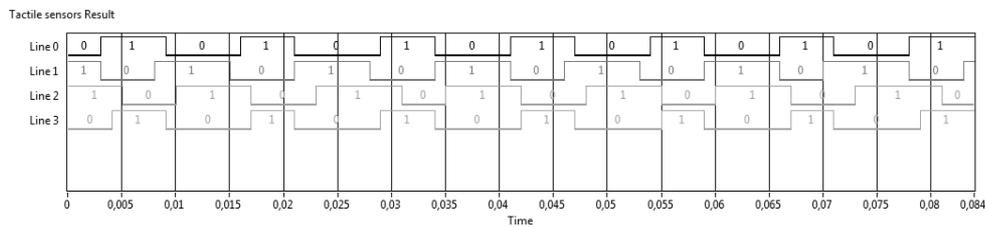
5. Results and discussion

A laboratory model with two hinge joints and attached pneumatic muscles was designed in order to test the control algorithm and system functionalities. A healthy subject equipped with special shoes with four pressure membranes and tactile sensors mounted under the heel and the toes part of the insole performs different trials of slow and normal level walking. Lower limbs movement was measured during walking using signals from sensors. A potentiometer is mounted on the hinge joint, coinciding with the axis of the ankle joint. Hinged joints are attached laterally of both ankles. The footswitches placed under the foot beneath the heel and the toe have been used to detect in real time the precise moments of heel strike (when the foot first touches the floor) and toe-off (when it takes off). 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.

A LabView virtual instrument is developed for visualization of the signals. The data from sensors were collected with multifunctional (DAQ) module (NI-USB-6211, National Instruments and LabVIEW). The graphic in Fig. 8a shows the kinematics of the ankle. Ankle angle rotation (potentiometer data in volts) is shown with the presence of peaks during the flexion or negative peaks



(a) Ankle angle rotation (potentiometer data in volts). Ankle L – recorded analogue signal for the left ankle; Ankle R – recorded analogue signal for the right ankle



(b) Tactile sensors results

Fig. 8. Visualization of human motion data (in LabView). Line0, Line1 – recorded digital signals from the switches mounted under the heel and the toes part of the left leg insole; Line2, Line3 – signals from the right leg

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 seen that the signals for left and right ankle joint have approximately the same values, but with opposite signs. The angle of rotation is measured relative to the value of 2 volts (i.e. the signal is shifted by 2 volts). 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 (Fig. 8b). 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.

Obtained signals are shown as a raw data. Further, these signals will be processed and visualized in a special program written in Matlab, for graphically displaying the angle of ankle rotation and different phases of walking.

It is obvious that the algorithm is applied in the same way for both legs. The pneumatic muscles possess a high power to weight ratio and can be coupled

directly without complex gearing mechanism. Due to the compressibility of air, a joint actuated with these pneumatic actuators shows a compliant behaviour, which can be positively employed to reduce shock effects. Joint compliance can be adapted while controlling position of the knee joint which enhances the possibilities of exploitation of natural dynamics.

6. Conclusion

The presented propulsion system for control of active ankle-foot orthosis integrates adaptive pneumatic system with artificial muscles and biomechanics based algorithms. Ankle joint actuation is realised with custom made pneumatic muscles supplied with compressed air by pneumatic membranes placed under the soles of the feet during normal level walking. The pneumatic system with artificial muscles is automatically modulated in order to optimize the heel-to-forefoot transition during the stance or the swing phase of walking. The data obtained from the sensors are used in every step from the control algorithm.

The proposed control device can be used in the new propulsion system with pneumatic artificial muscles to control the orthosis functionalities for ankle joint articulation, for assisting and lower limb rehabilitation.

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