Embedded System Design to Control Active Ankle Foot Orthosis

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***Abstract*-Active ankle foot orthosis is used to support or restore the complex ankle foot movements of the subjects with neurological or muscular pathologies. Neurological impairments are due to stroke, polio, multiple sclerosis, spinal cord injuries, and cerebral palsy. This paper aims at designing an embedded system for active ankle foot orthosis, which can be used in the case of hemiparesis subjects as a rehabilitation aid. The designed orthosis can aid both dorsiflexion and plantar flexion movements. The ATmega328 microcontroller is used to control the position of the actuator. The microcontroller generates pulse width modulation signals according to the outputs given from the tactile sensors. The sensors are mounted inside the soles of a healthy leg and the orthotic device. During each gait cycle the microcontroller estimates the position of the foot in relation to the gait phases based on the information obtained by the tactile sensors and the adaptation of the joint torque with the help of an actuator.**

Keywords-*Embedded System Design; Ankle Foot Orthosis; Orthotic; Gait; Tactile Sensor; Arduino Uno; Microcontroller*

I. INTRODUCTION

Human gait is defined as a manner of walking or moving on foot. It is a complex synergy of muscle coordination, timing, and balance. The efficiency and effectiveness of gait depends on joint mobility and muscle activity, which are both selective in terms of timing and intensity [1, 2]. The forces and motions generated during gait are attributed to three main functional tasks: weight acceptance, single limb support, and limb advancement. Weight acceptance and single limb support occur during stance phase when the foot is in contact with the ground, whereas limb advancement takes place during swing when the foot is off the ground. The ability to walk can be impaired by injuries, as well as numerous neurological and muscular pathologies [1, 3]. Neurological impairments are due to stroke, polio, multiple sclerosis, spinal cord injuries, and cerebral palsy. The lower limb impairments are frequently treated with ankle foot orthoses (AFOs) as a means of improving functionality during gait.

AFOs are orthotic devices worn externally on the foot to support or restore the motions of the complex ankle foot. AFOs can be grouped into passive, semi active and active device. Passive devices provide the resistance or support which is not changing in real-time. Semi-active devices use computer control to vary the compliance of the joint in real-time. Fully active devices have on-board or tethered sources of power, actuators to move the joint, sensors, and a computer or electronics to control the application of assistance during gait [1-5]. Our focus is on active AFO design.

Design considerations for the ideal active AFO must account for diverse functionality required to accommodate the many aspects of pathological gait. The core challenges are: i) a compact power source capable of day scale operation. ii) compact and efficient actuators and means of un-tethered power transmission capable of providing forces comparable to healthy individuals. iii) control schemes that effectively apply assistance during variety of functional tasks that an individual may encounter on a daily basis [1, 4].

In recent years, a number of research works has been carried out in the design of active AFO. A portable powered AFO was designed using pneumatic based actuator. Solenoid valves were used to feed the power to the pneumatic actuator but the drawback is that they were not able to provide sufficient amount of torque in the different phases of gait and motion control and the device consumes large amount of pneumatic power. The weight of the AFO is 3.1 kg [1]. A biofeedback AFO using Electromyography (EMG) signals is designed which provides only dorsiflexion assistance [2].The MIT Biomechanics group developed a powered AFO to assist drop foot gait [6]. The device consists of modified passive AFO. A series elastic actuator is added to allow the variation in the impedance of flexion/extension direction of ankle motion. The series elastic actuator used in this system is too heavy and power intensive. Each time the joint impedance needs to be adjusted to reduce the occurrence of slap foot. The AFO weights 2.6 kg and is tethered to an off board power supply [5, 6]. Arizona State University researchers have designed an AFO using pneumatic muscles as an actuator. The device is computer controlled which can be used for rehabilitation purpose. The weight of the device alone without adding the controller weight is 1.76 kg [7]. An ankle foot orthotic device has been developed which involves artificial pneumatic muscles which in turn are controlled by computer controlled air pressure based on acquired EMG signals [8]. Further this orthosis is improved by adding carbon fiber polypropylene shell, a metal hinge joint and two artificial pneumatic muscles. This orthosis is helpful in studying human walking biomechanics and assisting patients during gait rehabilitation after neurological injury. It can be used in gait laboratory or rehabilitation clinic where compressed air and electrical power is easily provided [8-9]. The powered AFO is designed using pneumatic artificial muscle as an actuator. The limitation of the design is that it is not readily portable due to the type of actuator used. It can be made portable using a micro air compressor [10]. Tactile sensors, voice-coil actuators position transducers and microcontroller are used in the control system of active AFO. The system controls the orthosis functionalities, records the data received from sensors during the gait, and transfers the recorded data to graphical user interface for visualization and further analysis [11-12]. A spring damper PD control is used to control the swing phase of an active AFO to assist drop foot gait [13]. It is a semiactive device. A portable knee brace rehabilitation device was designed which can be controlled through computer in real-time [14].There are no commercially available active orthosis which are compact, energetically autonomous that can provide both assistance and therapy during the wearer's everyday life [3,5].

The paper aims at design of an intelligent, untethered device for control of active AFO which can be used for assisting and rehabilitation in cases of injured ankle-foot complex mainly for hemiparesis subjects. Proposed AFO is with one degree of freedom in which foot segment is connected to the shank segment by a rotational joint. The joint is fitted with a servo motor attached laterally. The position control of the actuator is handled by pulse width modulation from an embedded system according to the outputs from force sensors mounted inside soles placed on the healthy leg and the orthotic device. During each gait cycle a microcontroller estimates the position of the foot in relation to the gait phases based on the information obtained by the force sensors and the adaptation of the joint torque with the help of an actuator in this case a motor. In monitoring mode data acquired from the sensors during human motion are transferred and noted, and are used to determine the threshold values. These threshold values are then used to obtain the limits for various phases of gait cycle and then the dorsi and plantar assistance by the motor is then given.

II. THEORY AND METHODS

The gait cycle is defined from the initial contact of the heel to the following heel contact as illustrated in Fig.1.The orange highlighted in the figure is the leg suffering from pathological gait [15].Every gait cycle is divided into two periods, stance and the swing. Stance is the term used to designate the entire period during which the foot is on the ground. Stance begins with initial contact i.e. heel strike. The word swing applies to the time the foot is in the air for limb advancement. Swing begins as soon as the foot is lifted from the floor i.e. toe-off condition.

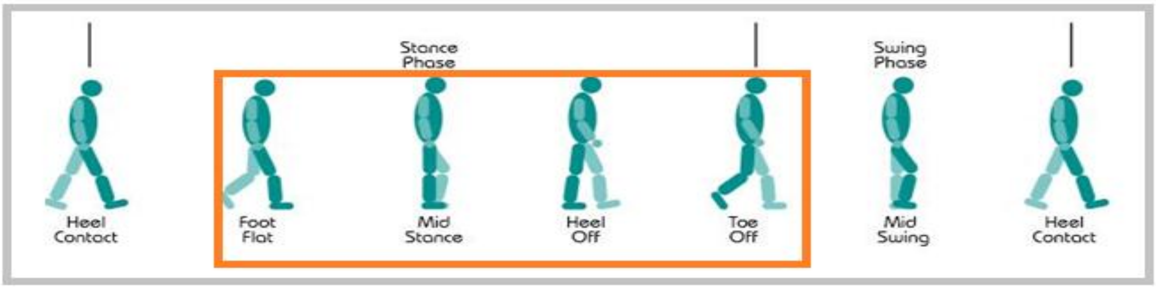


Fig.1 Human gait cycle showing stance and swing phases. The inner rectangle shows the gait phases suffering from leg pathology.

During normal gait, the ankle joint, shank, and foot play important roles in all aspects of locomotion including: motion control, shock absorption, stance stability, energy conservation, and propulsion [1].Ankle joint and foot involves a wide range of movements, mainly ankle joint allows only dorsiflexion and plantar-flexion movements. Inversion and eversion of foot is due to the movements that occur between tarsal bones. A simplified diagram of ankle joint in sagittal plane is considered, thus making the ankle to have one degree of freedom to do plantar-flexion or dorsiflexion is shown in Fig.2.

Weakness in the dorsiflexor and plantar flexor muscle groups is a key cause of impaired gait. Understanding muscle weakness and its effect on gait is essential to the proper design of orthotic devices that compensate for these deficiencies [16]. Muscle weakness can be neurological or muscular in origin and can be due to a multitude of pathologies. Common conditions that may result in muscle weakness include trauma, incomplete spinal cord injury, brain injury, stroke, multiple sclerosis, muscular dystrophy, and cerebral palsy. The ideal orthosis for the compensation of muscle weakness should be flexible enough to accommodate both plantar and dorsiflexor weakness.

Pathologies that afflict the function of the ankle dorsiflexor muscles affect gait in both swing and initial stance phases. Swing is affected by insufficient toe clearance due to weak or absent dorsiflexor muscles and results in a steppage type gait pattern that is commonly called foot drop. Steppage gait is a compensatory walking pattern characterized by increased knee and hip flexion during the swing phase to ensure that the toe clears the ground during walking. Weakness in the ankle plantar flexor muscle group primarily affects the stance phase of gait. From heel strike to middle stance, the ankle plantar flexors eccentrically contract to stabilize the knee and ankle and restrict forward rotation of the tibia. At the end of stance, the plantar flexors concentrically contract to generate torque that accelerates the leg into swing and contributes to forward progression. Weak ankle plantar flexors affect stability, particularly during single limb support. Individuals with impaired ankle plantar flexors compensate by reducing walking speed and shortening contralateral step length. Reduced walking speeds result in a corresponding reduction in torque needed for forward progression [1, 15, 16].

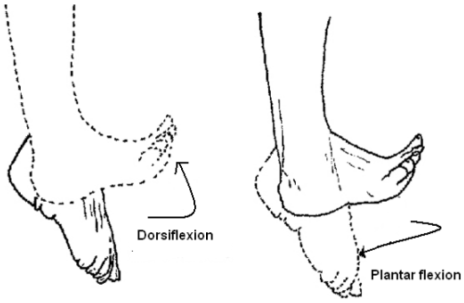


Fig.2 Ankle joint representing dorsiflexion and plantar flexion.

A. General Analysis

The active AFO will be designed to provide assistance during the following simplified functional gait tasks.

- Motion control of the foot at the start of gait cycle: The motion will be extension in the dorsiflexor muscle. This motion will cause the foot to move away from the shank segment in the lower human limb, under the knee. Start of the gait cycle is defined by heel strike. Such motion assistance will help in preventing foot-slap i.e. uncontrolled strike of foot on the ground while entering the stance phase.

- Motion assistance during terminal stance or pre-swing phase: This motion will be extension in the plantar flexor muscle. A propulsive force is required to assist the plantar flexor muscle to be able to lift the foot of the ground. This helps in preventing toe-drag. Thus the weak plantar flexor muscle is provided with external torque to do the required job of preventing toe-drag.

- Motion assistance for dorsiflexor control of foot during swing phase: This motion assistance is required by the dorsiflexor muscle to provide ground clearance for the foot while it is in swing phase. This motion involves flexion of the dorsiflexor muscle so that the foot moves closer to the shank segment or the shin of the lower limb. The torque is provided in anti-clockwise direction by the external actuator.

B. Event Triggering

The motion assistance to a patient suffering from pathological gait consists of either flexion or extension motions. The three functional gait tasks are identified on the basis of the triggering inputs obtained from the tactile sensors. These sensors are placed under the sole of both the healthy and the abnormal leg. The sensor inputs are used to control the actuator which is connected to the AFO. Each foot has two sensors, one under the ball and the other under the heel. Table.1 shows the event trigger for torque assistance. This table assumes that the right leg is the one having pathological gait and the left one corresponds to the healthy leg. SL1 and SL2 are the sensors for the left leg whereas SR1 and SR2 are the sensors for the right leg placed under heel and ball respectively as shown in Fig.3.

TABLE 1 EVENT TRIGGERING CHART

SL1	**SL2**	**SR1**	**SR2**	**CONTROL OBJECTIVE**	**TORQUE PROVIDED**
X	X	1	X	LOADING RESPONSE	DORSI
X	X	1	1	MID STANCE	NO TORQUE
1	0	0	1	TERMINAL STANCE	PLANTAR
1	1	0	0	SWING	DORSI

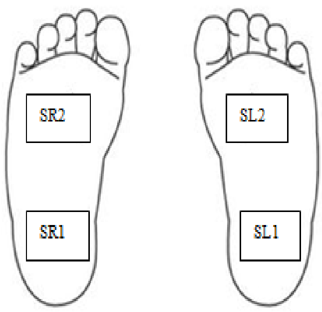


Fig.3 Sensor position

The heel of the foot and ball of the foot are the prominent areas of contact while walking and have been experimentally proved in many researches as maximum pressure bearing areas while walking or standing [4, 15]. Thus depending on the voltage output obtained from these sensors, the foot contact with the ground is determined. In the table.1 ‘X’ represents don’t care logic situation. A '1' represents logic high state and '0' represents logic low state. Depending on the combination of inputs obtained from all four sensors, the required motion assistance is provided to the orthosis..

C. Orthosis Design

The active AFO consists of sensor, control system and an actuator. In the proposed active AFO, Force Sensing Resistor, FSR 402 is used as a sensor to detect the force/pressure applied by the foot in different phases of gait. FSRs are two-wire devices as shown in Fig.4.

Fig.4 FSR 402 by Interlink Electronics

Fig.5 shows the sensor interface with the sole. FSR is optimal for use in human touch control of electronic devices such as automotive electronics, medical systems, and in industrial and robotics applications. It has a 13 mm diameter active area and is available in 4 connection options. They are robust polymer thick film (PTF) sensors that exhibit a decrease in resistance with increase in force applied to the surface of the sensor. The minimum and maximum forces are obtained from the sensor specifications as 0 g – 1000 g. Therefore any intermediate force can be obtained from a linear equation as:

$$\text{Force in grams} = \frac{1000\, g * ADC\ \ Value}{1023}$$

Fig.5 Sensor interface with the sole

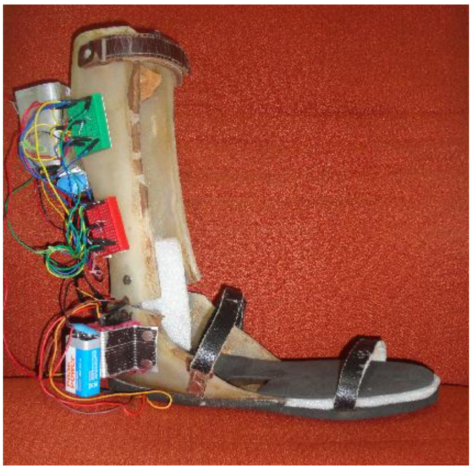


Fig.6 Side view of un- tethered Active ankle foot orthosis

Arduino Uno board was used as the controller. Arduino provides a number of libraries which make programming the microcontroller easier. Its small size saves space on the structure. The board has ATmega328 controller which can run with an operating speed of 20MHz and operating voltage 1.8-5.5V. The microcontroller has inbuilt ADC and PWM support. The special feature includes ADC noise reduction and extended standby mode.

High torque metal gear standard servo motor was selected as an actuator to replace the damaged muscle complex like a tendon in series with a muscle. The torque output of the motor at 4.8V is 1.4 Nm and at 6V is 1.6 Nm. Simulink model has been used [11, 12] in order to test and verify the torque output of the motor selected. 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. By varying the current flow though the coil the speed and torque of the motor can be varied. The motor was interfaced to the Arduino board and its control was tested.

The proposed active AFO design was made by taking a rigid, passive hard plastic structure. The side view of active AFO is as shown in Fig.6. The weight of the complete device is 1 kg. The complete circuit diagram showing Arduino interface with sensors, motors and power supply is shown in Fig.7. The design process of AFO is as follows:

1. Planning and marking of positions of various holes were made that were required to modify the structure to suit the dorsi and plantar angular requirements of 20 degrees each and then drilled. A rectangular opening was made to fit the motor accurately. Using the holes provided in the orthosis, motor mounting was done. Grinding was also done to improve the finishing of the structure.

2. Coupling of motor: a three-layered thick rounded rectangular strip was developed which was fixed to the shank but has holes for the motor to couple with. The other side was then screwed into the motor.

3. Two bread board circuits were developed to incorporate the sensor voltage divider circuits, each bread board containing the circuitry for two sensors.

4. Development of sole housing for the sensors was done. Two sensors were housed in between a pair of soles to protect the sensors from mechanical damage yet provide correct output when pressed.

5. The structure was developed by using housing for electronic components and motor. A program was written to test the motor to ensure that dorsi and plantar movements are working without any obstruction.

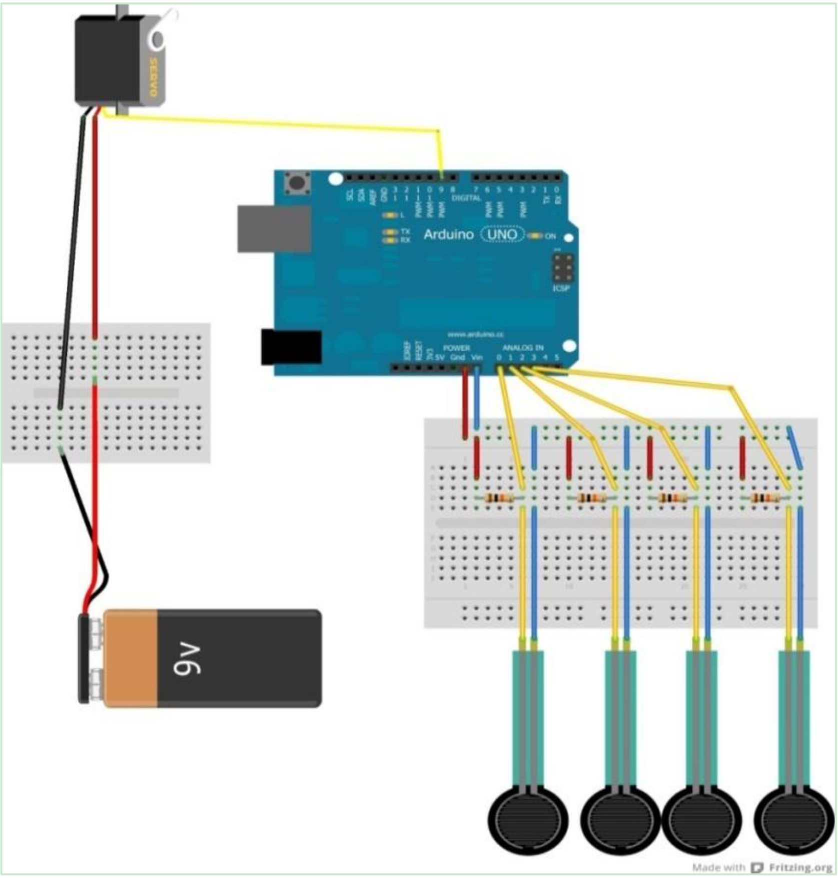


Fig.7 Complete circuit image showing wiring of servo motor, power supply and sensors using a breadboard to the Arduino Uno

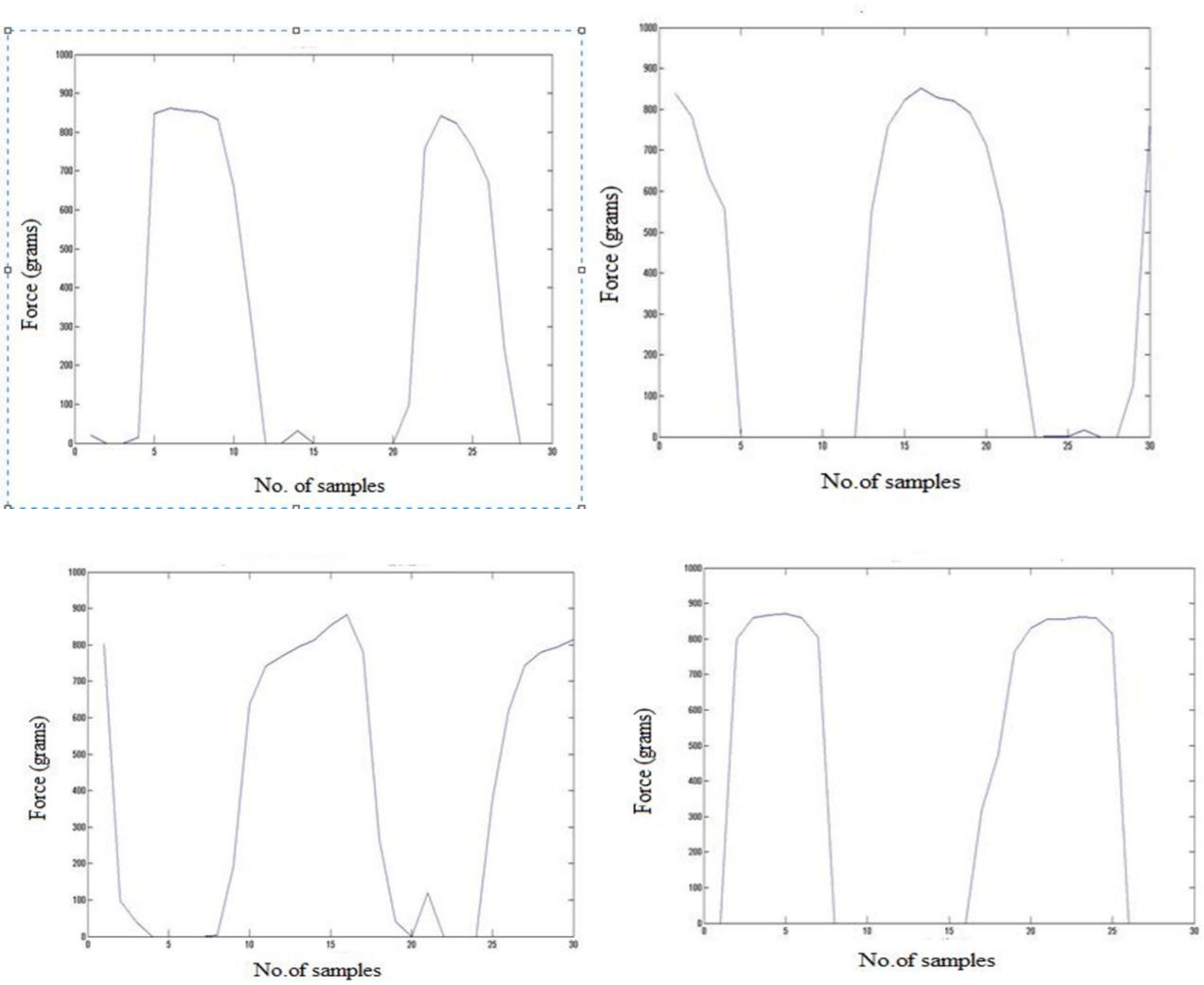
III. EXPERIMENTS

A. Threshold Level Monitoring and Testing

The two soles were connected with the four sensors each having two sensors on ball and heel of the sole. The sensors were interfaced with Arduino board and tested while walking. The Arduino was interfaced to MATLAB and graphs were generated for sensing the positions of ball and heel of foot. To determine the various threshold limits required for each of the four positions of the gait, the sensor values were tested.

The graphs for force against number of readings for two gait cycles for the four sensors are shown in Fig.8. It is observed that the graphs follow the cycle presented in Fig.1.and also the event triggering chart given in Table.1. Beginning from heel strike on the left foot, we can observe that the force for left ball is zero and force for left heel is maximum. At these points the right foot has maximum force at ball and decreasing force at heel. Then as the left foot is moving to stance, the force value at its ball increases and at the heel it decreases. Simultaneously, on the right foot, we can see the foot moving into swing, where force at both heel and ball are minimum, because there is negligible contact of sensor with the ground.

When the right foot goes into heel strike, force at heel shoots up to a maximum, the force value at ball still being minimum. As the left foot proceeds to toe-off, the force value at the ball increases to a maximum simultaneously decreases the force of the heel to a minimum. Then the right foot proceeds to its stance phase, where the force at its ball increases and force at its heel decreases. The whole process gets repeated for each gait cycle.



(a) (b)

Fig. 8 The force sensor reading for the (a) left foot and (b) right foot heel and ball simultaneously for the two gait cycles

B. Control Algorithm

The control algorithm involves determination of positions of foot during press on the various force sensors at different phases of gait cycle. The algorithm is based on the force sensor readings obtained as shown in Fig.8. Real time data has been obtained from MATLAB and used for determining thresholds. The threshold values of the tactile sensor readings are set up by taking number of trials with different subjects and the same is shown in Table 2. These were then used to actuate motor clockwise or anticlockwise to give plantarflexion or dorsiflexion assist respectively.

TABLE 2 THRESHOLD CALCULATIONS

THRESHOLD PARAMETER	FORCE VALUE (GRAMS)
LOW (Low_{tH})	100
MINIMUM (Min_{tH})	200
MEDIUM (Med_{tH})	400
MAXIMUM (Max_{tH})	700

The position control is handled by electronics according to the outputs from a set of two sensor arrays; tactile sensors SR incorporated in the foot part of AFO and SL in the insole of the healthy leg. The sensors were used to obtain data at various points in the gait cycle. Based on the inputs from the sensors SL and SR, the control outputs are generated throughout the stance and the swing phases of walking.

The control algorithm is shown in Fig.9.

Assuming,

LBall - sensor attached to the ball of left (healthy) foot

LHeel - sensor attached to the heel of left (healthy) foot

RBall - sensor attached to the ball of right (unhealthy) foot

RHeel - sensor attached to the heel of right (unhealthy) foot

Consider LOW_{th}, MIN_{th}, MED_{th}, MAX_{th} are the threshold values for low, minimum, medium and maximum respectively. When both LHeel and RHeel are with maximum threshold values, both feet are touching the ground. In this case foot must be perpendicular to shank. When left leg is almost in the stance phase and right leg is trying to lift above the ground i.e toe off position, provide plantarflexion movement. Similarly when there is less medium threshold values from right leg, provide dorsiflexion movement for the swing phase.

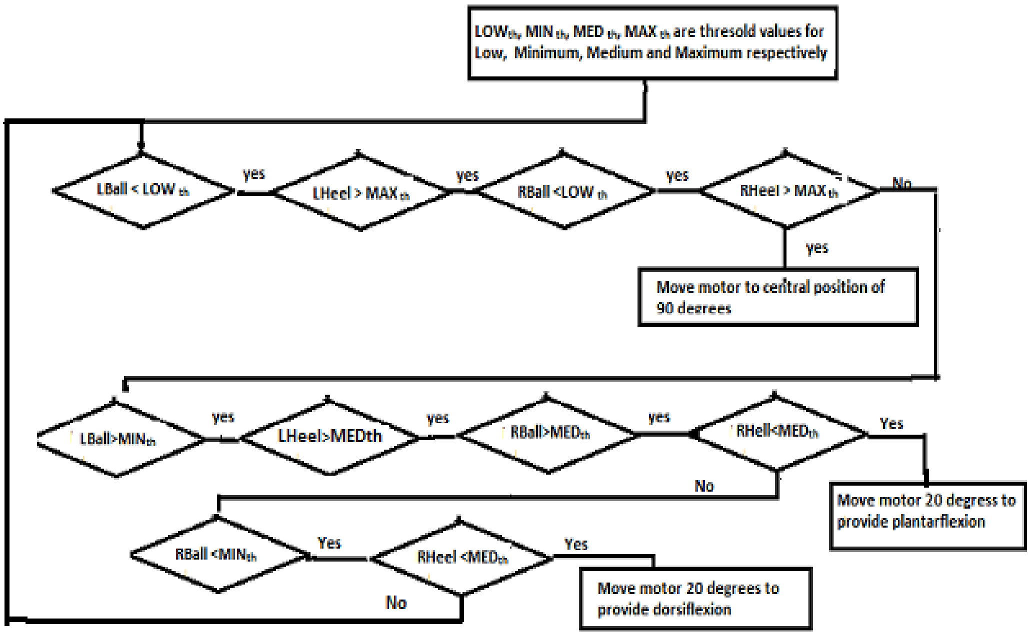


Fig. 9 The control algorithm

IV. RESULTS

The designed orthosis is tested on a 65Kg, normal subject. It was giving a sufficient amount of torque to move the foot in different phases of the gait. The treadmill testing of the active AFO is shown in Fig.10. The testing was done with a normal speed of walk 1 mt/sec. From the Fig.10, it is observed that the orthosis is simulating the movements of right leg based on the inputs from the left leg. The designed active AFO is able to give both plantar and dorsi assist to the disabled leg in different phases of the gait.

Position control servo motor is used as an actuator. The actuator is controlled with an Atmega 328 controller, run with a power source of battery. The design can be further improved by a speed control servo motor for the different walking speeds. In order to support body weight greater than 60-65kg, more than 1.4 Nm motor torque output is desired. But it may increase the weight of the whole system.

Fig.10 Treadmill testing of the device showing planar and dorsi assist for the right foot.

V. DISCUSSIONS

The results demonstrate that the proposed active AFO is capable of providing functional assistance for both dorsiflexor and plantarflexor impairments. However, these results also highlighted performance limitations associated with the hardware, as well as the use of a threshold for event detection. The structure needs to be modeled based on ergonomic study, stress tests need to be done to determine the maximum load it can bear and then needs to be fabricated. It should have some provisions through which it can fit patients of different leg sizes. This is a hurdle to generalizing the device. The weight of the structure is 1 kg and the material of the structure is plastic. If a composite material like carbon fibre or another polymer of plastic is made use of, not only will the device be lighter in weight but also provide better support. But carbon fibre is expensive, the use of it must be in specific areas of the orthosis only.We used position controlled servo motor which can operate with constant speed. Speed control is required when issue of varying walking speeds is addressed. The control algorithm we have used is based on different thresholds to determine the current position in the gait cycle. The system can be made efficient by using robust control techniques. Future work will continue to be directed towards transitioning this first generation system into a robust, viable device capable of meeting demanding system requirements during both rehabilitation (short-term goal) and as an effective daily-wear assist device (long-term goal).

VI. CONCLUSION

A portable, un-tethered active ankle foot orthosis with one degree of freedom is designed using embedded control system. The designed orthosis is light weight, battery operated having ATmega328 to control the actuator which is servo motor. Position controlled servo motor is used in this system. The position control of the actuator is handled by pulse width modulation from an embedded system according to the outputs from force sensors mounted inside soles placed on the healthy leg and the orthotic device. The sensors provide the feedback for the different phases of the gait to control the actuator. During each gait cycle the microcontroller estimates the position of the foot in relation to the gait phases based on the information obtained by the force sensors and the adaptation of the joint torque with the help of an actuator. In monitoring mode data acquired from the sensors during human motion are transferred and noted, and are used to determine the threshold values. These threshold values are then used to obtain the limits for various phases of gait cycle and then the dorsi and plantar assistance by the motor is given.

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