Chapter 5

Embedded System Design to Control Active AFO

5.1 INTRODUCTION

The ideal active AFO design must consider the 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 [4, 5,15]. 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 [1, 5].

A portable powered AFO was designed using pneumatic based actuator [4]. In the designed orthosis, 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. To overcome this limitation an embedded system for active AFO with one degree of freedom to control both dorsiflexion and plantar flexion movement is designed. DC servo motor is used as an actuator. The chapter discusses about the complete design of the orthosis and its control using ATmega328 microcontroller.
5.2 ORTHOSIS DESIGN

The active AFO consists of sensor, control system and an actuator. In the designed 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. FSR is optimal for use in human touch control of electronic devices such as automotive electronics, medical systems, and in industrial and robotics applications. FSRs are two-wire devices as shown in Figure 5.1. 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. It has a 13mm diameter active area and is available in 4 connection options. The minimum and maximum forces are obtained from the sensor specifications as 0 g – 1000 g.

Arduino Uno board is used as the controller. Arduino provides a number of libraries to 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 is selected as an actuator to replace the damaged muscle complex like a tendon in series with a muscle. The output of the motor at 4.8V is 1.4 Nm and at 6V is 1.6 Nm. Simulink model has been used [23, 24] 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 thought the coil the speed and torque of the motor can be varied. The motor is interfaced to the Arduino Uno board and its control has been tested. The complete circuit diagram showing the interface with Arduino Uno board with sensor and motor is shown in Figure 5.2.
The proposed active AFO design is made by taking a rigid, passive hard plastic structure. The side view of designed active AFO is as shown in Figure 5.3. The weight of the complete device is 1 kg. The design process is as follows:

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

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

3. Two bread board circuits are 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 is done. Two sensors are housed in between a pair of soles to protect the sensors from mechanical damage yet provide correct output when pressed.

5. The structure is developed by using housing for electronic components and motor. A program has been written to test the motor to ensure that dorsi and plantar movements are working without any obstruction.
Figure 5.2 Final circuit - Image showing wiring of servo motor, power supply and sensors using a breadboard to the Arduino Uno

Figure 5.3 Side view of un-tethered AFO
5.3 EVENT TRIGGERING IN THE GAIT CYCLE USING FORCE SENSORS

The motion assistance to a patient suffering from pathological gait consists of either flexion or extension motions. The functional gait tasks are discussed in chapter two, design requirements. Based on these gait tasks, triggering inputs are obtained from the tactile sensors. The gait cycle diagram is rewritten as in Figure 5.4, for clear understanding.

Figure 5.4 Human gait cycle showing stance and swing phases

<table>
<thead>
<tr>
<th>SL1</th>
<th>SL2</th>
<th>SR1</th>
<th>SR2</th>
<th>CONTROL OBJECTIVE</th>
<th>TORQUE REQUIRED</th>
</tr>
</thead>
<tbody>
<tr>
<td>X</td>
<td>X</td>
<td>1</td>
<td>X</td>
<td>LOADING RESPONSE</td>
<td>DORSI</td>
</tr>
<tr>
<td>X</td>
<td>X</td>
<td>1</td>
<td>1</td>
<td>MID STANCE</td>
<td>NO TORQUE</td>
</tr>
<tr>
<td>1</td>
<td>0</td>
<td>0</td>
<td>1</td>
<td>TERMINAL STANCE</td>
<td>PLANTAR</td>
</tr>
<tr>
<td>1</td>
<td>1</td>
<td>0</td>
<td>0</td>
<td>SWING</td>
<td>DORSI</td>
</tr>
</tbody>
</table>
The tactile sensors are placed under the sole of both the healthy and the abnormal leg. Each foot has two sensors, one under the toe and the other under the heel. The trigger table shown in 5.1 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. The position of sensors is shown in Figure 5.5.

As in the Table 5.1, the ‘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. The sensors used are called force sensors, whose resistance decreases as the amount of force on them increases. 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 [2, 16, 66]. Thus depending on the voltage output obtained from these sensors, we can determine which part of the foot is in contact with the ground.

### 5.4 GAIT ANALYSIS USING TACTILE SENSORS

The tactile sensor placement is as shown in Figure 5.6. Figure 5.7 and Figure 5.8 show the graphs of left heel and ball respectively, for two gait cycles when sensors are placed under left foot. Figure 5.9 and Figure 5.10 show the graphs of right heel and ball respectively, for two gait cycles when sensors are placed under right foot. It is clearly observed that the graphs follow the cycle presented in Figure 5.4. Beginning from heel strike on the left foot, it
is observed that the graph for left ball is zero and graph for left heel is maximum. At these points the right foot has maximum value at ball and decreasing value at heel. When the left foot is moving to stance, the value at its ball increases and at the heel it decreases. Simultaneously the right foot is moving into swing, where values at both heel and ball are minimum. This is due to negligible contact of sensor with the ground. When right foot goes into heel strike, the value at heel shoots up to a maximum, the value at ball is still being minimum. At this point, the left foot has proceeded to toe-off, thereby increasing the value at the ball to a maximum and decreasing the value of the heel to a minimum. This whole process gets repeated for each gait cycle.

![Tactile sensor interface with the sole.](image)

Figure 5.6 Tactile sensor interface with the sole.

![ADC output graph for ball of the left foot](image)

Figure 5.7 ADC output graph for ball of the left foot
Figure 5.8 ADC output graph for heel of the left foot

Figure 5.9 ADC output for graph for ball of the right foot
5.5 CONTROL ALGORITHM

The graphs for force against number of readings for two gait cycles for the sensors of left foot and right foot are shown in Figures 5.11 and 5.12 respectively. According to the force calculations from the sensor specifications of FSR 402, the range for force varies from 0 – 1000 g. Therefore any intermediate force can be obtained from a linear equation as [64]

\[
\text{Force in grams} = \frac{1000 \cdot ADC_{\text{value}}}{1023} \quad (5.1)
\]

The force values are proportional to the ADC value and hence we obtain the similar graphs as ADC output for the force data.
Figure 5.11 Force sensor readings of the left foot (a) heel (b) ball
Figure 5.12 Force sensor readings of the right foot (a) heel (b) ball
The control algorithm involves determination of positions of foot during press on the various force sensors at different times of gait cycle. The algorithm follows the event triggering chart shown in Table 5.1. Real time data is obtained from MATLAB and used for determining thresholds. These data are then used to actuate motor clockwise or anticlockwise to give plantarflexion or dorsiflexion assist respectively.

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 5.2. Considering $\text{Low}_{TH}$, $\text{Min}_{TH}$, $\text{Med}_{TH}$, $\text{Max}_{TH}$ are the threshold values for low, minimum, medium and maximum respectively.

<table>
<thead>
<tr>
<th>THRESHOLD PARAMETER</th>
<th>FORCE VALUE (grams)</th>
</tr>
</thead>
<tbody>
<tr>
<td>LOW $\text{(Low}_{TH}$)</td>
<td>100</td>
</tr>
<tr>
<td>MINIMUM $\text{(Min}_{TH}$)</td>
<td>200</td>
</tr>
<tr>
<td>MEDIUM $\text{(Med}_{TH}$)</td>
<td>400</td>
</tr>
<tr>
<td>MAXIMUM $\text{(Max}_{TH}$)</td>
<td>700</td>
</tr>
</tbody>
</table>

The control algorithm is as follows: 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

**IF** $(\text{LBall}<\text{Low}_{TH}$ AND $\text{LHeel}>\text{Max}_{TH}$ AND $\text{RBall}<\text{Low}_{TH}$ AND $\text{RHeel}>\text{Max}_{TH}$) *(This means that both feet are in stance phase, hence no assist should be given)*

**THEN** move motor to central position of 90 degrees.

**IF** $(\text{LBall}>\text{Min}_{TH}$ AND $\text{LHeel}>\text{Med}_{TH}$ AND $\text{RBall}>\text{Med}_{TH}$ AND $\text{RHeel}<\text{Med}_{TH}$) *(This means that left foot is almost in stance phase and right foot is in toe off phases)*

**THEN** move motor anti-clockwise to provide 20 degree downward assist
IF (RBall<Min\text{th} AND R\text{Heel}<Med\text{th}) (This means that the right foot is in the air, i.e., swing phase) THEN move motor clockwise to provide 20 degree upward assist

5.6 CONTROLLER TESTING AND VERIFICATION

The designed orthosis is tested on a 65Kg, normal subject. The sufficient amount of torque has been provided by the orthosis to move the foot in different phases of the gait. The treadmill testing of the active AFO is shown in Figure 5.13. The testing is done with a normal speed of walk 1 mt/sec. Here left foot assumed to be normal and right one is disabled. 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 i.e motion assistance during initial stance, motion control during terminal stance and motion assistance/ control during pre-swing. Hence justifying the design requirements with general analysis discussed in chapter 2, section 2.4.

Figure 5.13 Treadmill testing of the device showing plantar and dorsi assist for the right foot.

5.7 CONCLUSION

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 modelled 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 being expensive, the use of it must be in specific areas of the orthosis only. The control algorithm used in the design is based on different thresholds to determine the current position in the gait cycle. It is an open loop control system. Easy to implement and stable so far system concerned. The system can be made efficient by using robust control techniques and feedback can be added to adjust disturbances if any.