Chapter 2

Background Theory

2.1 INTRODUCTION

In the previous chapter basic structure of the research work is described. The motivation to undertake this project is explained where not only the importance of the work is justified but also the shortcomings of previous work in the same field have been mentioned. The primary objective of the work is described, and then the emphasis of the targeted result is laid. This chapter concentrates primarily on the literature survey done prior to the start of control system design. The significance of the project title is discussed in the section 2.2. The section 2.3 gives the literature survey which consists of the background theory required for understanding this project work. It also consists of the findings obtained upon reading various journal articles on state of the art on AFO design. The outcome of the literature review has been summarized which ultimately lays the ground work in determining the specific areas of research and improvement along with the sequence of tasks that has to be followed to achieve our objectives is also given in section 2.3. The general analysis of the pathological gait is done to find the requirement of active AFO assistance is given in section 2.4. Section 2.5 briefs about Electromyography signal and its applications. This chapter also presents the derivation involved in obtaining the mathematical modelling of the actuator in section 2.6.

2.2 SIGNIFICANCE OF RESEARCH PROJECT TITLE

The thesis title has been reproduced here for the reader’s convenience – “Embedded control system for Active Ankle Foot Orthosis”. Each word has a separate significance in defining the intent of the research work. The phrase “active ankle foot orthosis”, means the system being designed is actively and continuously controlled by an embedded electronic
system which controls both extension and flexion of the foot muscles around the ankle. On the contrary passive AFO system would control only the flexion of the foot muscle but not the extension. Even in the flexion, the motion is controlled due to mechanical forces alone and no role is played by any closed loop control design. Thus without any feedback these systems just act on previous calibration done before donning the orthosis.

2.3 GAIT ANALYSIS

Before beginning a technical discussion on the requirements and constraints for designing the controller, a brief introduction to normal and pathological human gait is necessary and same is explained herein.

Gait analysis is often defined as the study of human walking; typically involving computerised and instrumented measurement of the movement patterns that make up walking [2, 7]. It is gaining acceptance as a clinical tool for the investigation of complex gait disorders to plan for the treatments. Gait Analysis can reveal the timing and pattern of activation of muscles and joints, of body segment motions, and the forces that act on them. Analysis of a person’s walking pattern by phases more directly identifies the functional significance of the different motions occurring at individual joints. The phases of gait also provide a means of correlating simultaneous actions of the individual joints into patterns of total limb function. This is a particularly important approach for interpreting the functional effects of disability. The relative significance of one’s joints motion compared to the others varies among the gait phases. Also a posture that is appropriate in one gait phase would signify dysfunction at another point in the stride, because the functional need has changed. As a result, both timing and joint angle are very significant. This latter fact adds to the complexity of gait analysis [7-9]. Understanding human movement is essential when developing systems capable of assisting the human body [1].

2.3.1 Normal Human Gait

Walking is the body’s natural means of moving from one location to another. It is also the most convenient means of travelling short distances. Walking is a complex activity because it is dependent on a series of interactions between two multi-segmented lower limbs
and total body mass. Walking uses a repetitious sequence of limb motion to move the body forward while simultaneously maintaining stance stability.

As the body moves forward, one limb serves as a mobile source of support while the other limb advances itself to a new support site. Then the limbs reverse their roles. The transfer of body weight from one limb to the other takes place when both the limbs are in contact with the ground. This series of events is repeated by each limb, one by one, until the person’s destination is reached. A single sequence of these functions performed by one limb is called a Gait Cycle [2]. With one action flowing smoothly into the next, there is no specific starting or ending point. Because the moment of floor contact is the most readily defined event, this action has been selected as the start of the gait cycle. Normal person would initiate the floor contact with their heel. However, as explained later, pathological human gait may not necessarily include heel contact at the beginning of the gait cycle.

2.3.2 Gait cycle division

Each 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. 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. Figure 2.1 illustrates the different phases in a gait cycle of a single limb with much clarity. The internal rectangle in the picture highlights the gait phases of leg suffering from pathology.

![Figure 2.1 Human gait cycle showing stance and swing phases](image-url)
The stance phase is further subdivided into four sub-phases: loading response, mid-stance, terminal stance, and pre swing. Swing phase is divided into three periods: initial swing, mid-swing, and terminal swing. The beginning and ending of each period are defined by specific events.

The loading response marks the beginning of the stance phase and the end of swing phase for the limb. Here the initial contact with the ground is made i.e. heel contact. Mid stance is the phase where the entire weight of the body is balanced on a single limb. The terminal stance marks the lifting of foot heel off the ground; however the toes are still in contact with the surface. Pre-swing is the moment when the toe is just about to be lifted off the ground so that the limb can enter the next phase i.e. swing. While in swing phase the leg simply moves from previous location to the next location while the body weight is being balanced on the other limb [1].

2.3.3 Pathological Gait

Pathological or abnormal gait is characterized by drop foot which is the inability of an individual to lift their foot because of reduced or no muscle activity around their ankle. The major causes of drop foot are severing of the nerve, stroke, cerebral palsy and multiple sclerosis. There are two common complications from drop foot. First, the individual cannot control the falling of their foot after heel strike, so that it slaps the ground on every step. This is referred to as slap foot. The second complication is the inability to clear their toe during swing. This causes the person to drag their toe on the ground throughout swing. This is referred to as toe drag [10].

2.3.4 Ankle foot joint

In daily life ankle joint plays a very important role in balancing the gait. The formation of ankle joint is due to joining of distal ends of tibia and fibula with the talus. As it is a typical hinge joint, the capsule is thin anteriorly and posteriorly, which in turn strengthens the sides of special ligaments. Ankle joint and foot involves a wide range of movements, mainly ankle joint allows only dorsiflexion and plantarflexion of the ankle foot joint. Inversion and eversion of foot is due to the movements that occur between tarsal bones. Deltoid ligaments lie on the medial side which withstands the eversion of foot. The posterior talofibular ligaments and calcaneofibular ligament are responsible for limiting inversion of foot [11].
simplified model of ankle joint in sagittal plane is considered, thus making the ankle to have one degree of freedom to do plantarflexion or dorsiflexion is shown in Figure 2.2. It is due to majority of ankle movement occurred in sagittal plane in normal walking [12].

![Figure 2.2 Ankle joint with dorsiflexion and plantarflexion movements](image)

Figure 2.3 shows the lower leg muscles. The dorsiflexor muscle groups are situated anterior to the ankle joint and include the tibialis anterior, extensor digitorum longus, and extensor hallucis longus [2]. 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 [13]. The plantarflexor muscles situated posterior to the ankle joint are comprised of the gastrocnemius, soleus, and the peroneal and posterior tibial muscles [2]. From heel strike to middle stance, the ankle plantarflexors eccentrically contract to stabilize the knee and ankle and restrict forward rotation of the tibia. At the end of stance, the plantarflexors concentrically contract to generate torque that accelerates the leg into swing and contributes to forward progression. Weak ankle plantarflexors affect stability, particularly during single limb support [13]. Individuals with impaired ankle plantarflexors compensate by reducing walking speed and shortening contralateral step length. Reduced walking speeds result in a corresponding reduction in torque needed for forward progression. Impaired individuals may maintain a fast walking pace by using their hip flexors to compensate for weak plantarflexor muscles [14].
2.4 LITERATURE REVIEW

AFOs can be used to ameliorate impact of impairments to the lower limb neuromuscular motor system that affect gait [4, 5]. The standard AFO is a rigid polypropylene structure that prevents any ankle motion. AFOs can be grouped in to passive, semi active and active device. Passive devices provide the resistance or support which are not changing in real time. Semi-active devices use computer control to vary the compliance of the joint in real-time. 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 [4, 5, 16]. The active AFOs provide net power to the ankle for use in both motion control and propulsive assistance. Emerging technologies provide a vision for fully powered, un-tethered AFOs. However, stringent design requirements such as light weight, small size, high efficiency and low noise present significant engineering challenges, that must be met before such devices are realized.

Several studies have analyzed different ideas for designing actively powered orthotic devices that could be of more benefit, it uses hydraulic and pneumatic devices or DC motors for assisting individuals with stroke or drop foot [1]. An early active ankle orthosis was presented in 1981 by Jaukovic at the University of Titograd in the former Yugoslavia [1]. The
device consisted of a dc motor mounted in front of the wearer’s shin that assisted in the flexion/extension of the ankle. The orthosis was controlled based upon the information from foot switches in the soles. A spring damper PD control is used to control the swing phase of an active AFO to assist drop foot gait [12]. It is a semi-active device.

In 2005 the Human Neuromechanics Laboratory at the University of Michigan has developed a number of active orthoses. These orthoses are primarily designed for lower leg, with both ankle-foot and knee-ankle-foot devices [1]. In these devices carbon fiber and polypropylene shells are custom–built for each subject, eliminating the need for mechanically complex adjustment mechanisms. In the same year researchers in the Department of Mechanical Engineering and Physical Therapy at the University of Delaware have also proposed a design of an active AFOs that adds power to the wearer in both the flexion/extension and inversion/eversion directions.

Researchers have presented a novel design of an active AFO with two “spring over muscle” actuators attached to the left and right sides of the foot under toes, forming a tripod with the heel [1]. These actuators are pneumatic muscles with internal spring tending to extending the muscle, enabling force to applied in both plantar and dorsiflexion directions.

A portable powered AFO was designed using pneumatic based actuator. Soleniod 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 [4].

A biofeedback AFO using Electromyography (EMG) signals is designed which provides only dorsiflexion assistance to the ankle joint when the user is unable to voluntarily provide the same motion [17]. This AFO is used to assist the user for the following circumstances. (1) When the user generates a voluntary EMG signal in an attempt to raise their ankle, but no dorsiflexion motion is registered during the initial swing phase and (2) when the user enters the initial swing phase and no voluntary EMG signal is detected. The second circumstances serve as a fail-safe procedure to prevent unstable gait patterns caused by unwanted contact of the toe and foot resulting from drop-foot. It is tethered and used for rehabilitation.
The Biomechanics group at Massachusetts Institute of Technology, Cambridge, U.S., developed a powered AFO to assist drop foot gait [18]. The device consists of modified passive AFO with the addition of a series elastic actuator to allow the variation in the impedance of flexion/extension direction of ankle motion, controller based on ground force and angle position data. 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 [1, 18].

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 [19]. 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 [19, 20]. 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 [21]. The AFO has a total weight of 1.6kg excluding the off board computer and air compressor.

Arizona state university researchers have designed an AFO with two “spring over muscle” actuators called as robotic tendon, attached to left and right sides of the foot under toes, forming a tripod with the heel [22]. These actuators are essentially pneumatic muscles with an internal spring tending to extend the muscle, enabling force to be applied in both plantar and dorsiflexion directions. The design used a 0.95kg tendon that required 77 W of power to produce a torque comparable to a healthy individual during level walking. This device is computer controlled and mainly used for rehabilitation purpose.

Tactile sensors, voice-coil actuators position transducers and microcontroller are used in the control system of a laboratory model of active AFO [23-24]. 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. The maximum torque support by this orthosis during swing phase is 1.4214 Nm [23].
A portable knee brace rehabilitation device was designed which can be controlled through computer in real time [25]. The ToeOFF(tm) is a compact, lightweight AFO that allows a wide range of ankle motion, while providing a dorsiflexion moment.

After performing an extensive review of various scientific journals, the problem at hand has been determined. The current orthosis used for rehabilitation purpose provide a crude mechanism for preventing toe drag i.e. dorsiflexion assistance. However since the mechanism does not replicate the normal human gait, it can’t help in restoring the body’s natural walking cycle. Thus it is mandatory to develop a control structure which will enable a smooth transition or angular motion of leg, almost duplicating the normal human gait which will inevitably lead to restoring of natural muscle activity. The survey also helped in determining the design requirements for the control system to counter act all the problems encountered in pathological gait. This shall be explained in the next section.

The design considerations for the ideal AFO must account for the diverse functionality required to accommodate the many aspects of gait that can be affected by the leg pathology. It also must be compact and light weight to minimize the energetic impact to the wearer. These requirements illustrate the great technological challenges facing the development of untethered, powered AFOs for daily wear.

2.5 DESIGN REQUIREMENTS

The Active ankle foot orthosis is 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.
According to Figure 2.4, it is inferred that the clockwise direction motion is being controlled. This motion although leads to the neutral position (foot resting on ground) but is still controlled by the dorsiflexor muscle instead of the plantarflexor muscle.

- Motion Assistance during terminal stance or Pre-swing phase: This motion will be an extension in the plantarflexor muscle. A propulsive force is required to assist the plantarflexor muscle to be able to lift the foot of the ground. This helps in preventing toe-drag. Thus the weak plantarflexor muscle is provided with external torque to do the required job of preventing toe-drag. The illustration shown in Figure 2.5 clearly explains this concept. The torque is provided in the clockwise direction and the foot is lifted in upward direction.

- 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 such 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 as shown in Figure 2.6.
2.6 ELECTROMYOGRAPHY SIGNAL

Electromyography (EMG) is a technique of acquiring and recording an electrical activity generated by muscles. Electrical activity is due to functional variations in muscle fibre membranes [26]. There are many applications for the use of EMG. EMG is used clinically for the diagnosis of neurological and neuromuscular problems. It is used diagnostically by gait laboratories and by clinicians trained in the use of biofeedback or ergonomic assessment. EMG is also used in many types of research laboratories, including those involved in biomechanics, motor control, neuromuscular physiology, movement disorders, postural control, and physical therapy.

Surface electromyography (sEMG) signals, which are collected from the skin covering muscles contain rich information that can be used to recognize neuromuscular activity in a non-invasive manner [27]. Raw EMG offers us valuable information in a particularly useless form. This information is useful only if it can be quantified. Various signal-processing methods are applied on raw EMG to achieve the accurate and actual EMG signal. During the EMG signal processing only positive values are analysed. When half-wave rectification is performed, all negative data is discarded and positive data is kept. The absolute value of each data point is used during full-wave rectification. Usually for rectification, full-wave rectification is preferred. Precise detection of discrete events in the sEMG is an important issue in the analysis of the motor system [26, 27].

sEMG has undergone a vast research for many decades by many researchers in the fields of medical and biomechanical, mainly to understand the internal behaviour of muscles when they are activated. The signal is widely used as a suitable means to have access to the physiological processes involved in producing joint movements. The research in recent years resulted in obtaining the relationship between single muscles and their complicated movements involved in daily activities of human body [28]. Most of the recent studies mainly
focused on examining disabilities and their effects on EMG activity [29, 30]. The information extracted from the EMG signal can be used to control rehabilitation devices or to study the biomechanics and motor control of musculoskeletal system during different movements of the upper and lower extremities. In the case of biomechanical research, important information about the natural and pathological functioning of the neuromuscular system can be obtained in order to assess posture and movements in able-bodied and disabled persons. To detect the muscle contraction levels the raw EMG is compared with the fixed threshold. But the probability of detection is very low since there is only one parameter to compare and it is suitable only when there is a good signal to noise ratio (SNR) [31-33].

2.7 MATHEMATICAL MODELLING OF DC MOTOR ACTUATOR

A differential equation governing the input and output of the DC motor dynamics has been used, which is then combined with a biomimetic foot model to compute the torque requirements of the ankle foot movement. The DC motor is modelled as a simple RL circuit. Also the back EMF generated by the movement of the armature of the DC motor in permanent magnetic field has been taken into account. Motor torque constant and back EMF constant have been included in the derivation along with the coefficient of viscous friction of the rotor. The modelling as will be shown in the following paragraphs, has followed a generalized pattern taking into account almost all physical constraints. However discrepancies shall always be there between the real motor and its mathematical counterpart. For design purposes the motor parameters were obtained from the datasheet of PITTMAN-MOTORS. The motor is a DC brush motor – ID33904. Figure 2.7 shows the schematic representation of the DC motor.
A generic model of a DC motor includes two windings; a stationary field winding on the stator and a second winding for the rotating armature. The speed of the motor can be controlled by varying either the field current or the armature current. Its torque to armature current relationship is given in Figure 2.7 where, $R_A$ is the series resistance of the DC motor, $L_A$ denotes the inductance of the armature coil, and $V_B$ represents the back emf. $V_M$ is the voltage applied across the terminals of the DC motor. $I_A$ is the current flowing through the armature coils.

Most modern DC motors are somewhat different in construction. The field winding is replaced with two or more powerful rare-earth magnets on the stator. Since the field strength of these motors is constant, they can only be controlled by varying the armature current, $I_A$. In a permanent magnet motor the output torque, $T_M$, is directly proportional to the armature current, $I_A$ as shown in Figure 2.8. The constant of proportionality is referred to as the torque constant of the motor and is represented by $K_T$.

The armature current, $I_A$, can be expressed in terms of the applied motor voltage, $V_M$, by applying Kirchhoff’s Voltage Law (KVL) to the armature circuit. As with most windings, the impedance of the armature can be represented by a resistive component, $R_A$, in series with an inductive component, $L_A$. 

![Figure 2.7 DC motor electrical equivalent [31].](image)
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$$V_M = (I_A \times R_A) + \left(L_A \times \frac{dI_A}{dt}\right) + V_B$$  \hspace{1cm} (2.1)

The voltage, $V_B$, represents the back Electromotive Force (back EMF). Most permanent magnet DC motors can also operate generators. In other words, if the motor shaft is rotating, a voltmeter connected across the motor terminals will indicate a voltage that is proportional to the angular velocity of the motor shaft, $\omega_M$. This “generator phenomenon” occurs even when the motor is operating in the normal manner (i.e., as a motor). According to Lenz's law, the polarity of the induced back EMF voltage, $V_B$, is such that it always opposes the applied voltage, $V_M$. The net result is that a DC motor with no load (and in the absence of friction) will reach a maximum steady-state angular velocity corresponding to the operating point where the induced voltage, $V_B$, completely offsets the applied voltage, $V_M$.

The back EMF voltage $V_B$, can be expressed as a function of the angular velocity of the motor, $\omega_M$ as follows:

$$V_B = Ke \times \omega_M$$  \hspace{1cm} (2.2)

The back EMF constant, $Ke$, is often numerically identical to the torque constant of the motor, $K_T$ (at least in the SI system of units). This is not surprising since both terms are related to the geometry of the motor, the field strength, and the length of wire moving through the magnetic field. All three terms are constant in a permanent magnet DC motor. An expression relating the armature current, $I_A$, to the applied motor voltage, $V_M$, and the angular velocity of the motor, $\omega_M$, can be found by substituting $Ke \omega_M$ for $V_B$ in the armature circuit equation 2.2.

$$V_M = (I_A \times R_A) + \left(L_A \times \frac{dI_A}{dt}\right) + Ke \times \omega_M$$  \hspace{1cm} (2.3)

If the armature current is initially zero (i.e., no initial conditions), this equation can be written in the Laplace domain as given in equation 2.4.

$$V_M = (sI_A + (L_A \times s) + Ke \times \omega_M$$  \hspace{1cm} (2.4)
This equation can be expressed as a transfer function as given in Figure 2.9

![Figure 2.9 Current transfer function block](image)

By including the torque constant of the motor, $K_T$, the motor torque, $T_M$, can be related to the applied voltage, $V_M$ as shown in Figure 2.10

![Figure 2.10 Torque transfer function block](image)

In order to complete the transfer function of the motor, a means of relating torque to angular velocity must be found. This can be accomplished by considering the mechanical model of the motor. The torque generated by the motor accelerates the armature of the motor as well as any additional load inertia on the motor shaft. Some of the torque also goes towards overcoming friction. In order to maintain a linear system model, only the viscous motor friction will be considered for the time being. For a mechanical system undergoing pure rotational motion, Newton’s second law states that the sum of the applied torques is equal to the product of the mass moment of inertia, $J_G$, and the angular acceleration of the body, $\alpha$.

In the case of the DC servomotor, $J_G$ is equal to the sum of the mass moment of inertia of the motor armature, $J_m$, and the load, $J_l$. The net torque generated by the motor is equal to
the motor torque, $T_M$, minus the rotational viscous friction. The rotational viscous friction associated with the motor is proportional to the motor angular velocity, $\omega_M$. 

$$T_M - f_m \omega_M = (J_m + J_L) \alpha m$$

(2.6)

where, $f_m$ is the motor viscous damping coefficient. Replacing the angular acceleration of the motor with the rate-of-change of angular velocity yields:

$$T_M - f_m \omega_M = (J_m + J_L) \frac{d\omega_M}{dt}$$

(2.7)

Assuming zero initial conditions, this equation can be written in the Laplace domain as follows:

$$T_M - f_m \omega_M = (J_m + J_L) s \omega_M$$

(2.8)

The transfer function relating the motor angular velocity, $\omega_M$, to the motor torque, $T_M$, can be derived as given in equation 2.9 and corresponding model is shown in Figure 2.11.

$$\frac{\omega_M}{T_M} = \frac{1}{(J_m+J_L)s+f_m}$$

(2.9)

![Figure 2.11 Torque and Angular velocity relation](image)

The motor transfer function can now be completed as shown in Figure 2.12:

![Figure 2.12 Final Actuator Transfer Function model [34]](image)
2.7 CONCLUSION

The theory behind the design of active AFO is discussed in this chapter. The literature review gives the work done so far in this field and also their merits and demerits. The same gives a platform to begin the research work. The general analysis gives the design problem to find the various gait phases wherein there is requirement of torque control. Mathematical modelling of DC motor actuator is discussed. Before starting with actual AFO design and its control system it is very much necessary to understand biomechanics. The study of muscle activity is done using electromyography (EMG) which is discussed in the next chapter following prototyping of active AFO design using EMG signal.