DESIGN AND CONTROL OF ACTIVE KNEE REHABILITATION ORTHOTIC DEVICE (AKROD)

A Thesis Presented

by

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ABSTRACT

Patient populations with stroke or neurological disorders who have lost their walking capabilities regain ambulatory motor control functioning by undergoing physical rehabilitation. Conventional techniques are labor intensive and often require one-to-one administration, thereby making it time consuming for the therapist. This in turn leads to increased costs and reduced training duration. Research has shown that patient can improve their muscle strength and movement pattern by practicing locomotor related activities. With the advancement of robotic technologies in the recent years has led to the development of automated gait training devices to enable and assist patient to recover their motor skills. One such set of treadmill-based devices provides the means for intense rehabilitation but is very expensive and require large operational space.

Presented in this thesis a low cost portable device called Active Knee Rehabilitation Orthotic Device (AKROD). AKROD is designed and developed for gait rehabilitation of patient post-stroke and is targeted towards individuals who have regained a certain level of motor control but not yet fully attained normal walking capabilities. AKROD functions by providing active assistance to the patient’s lower limbs in order to reinforce the desired trajectory in terms of knee position and knee moment along the gait cycle. The proposed system allows for the normal movement of the knee joint along the sagittal plane to allow patient advancement and at the same time incorporates techniques to prevent hyperextension often displayed by these individuals.

A novel actuation mechanism called the Gear Bearing Drive (GBD) developed at Northeastern University is used to drive the AKROD. The Gear Bearing Drive is a
compact and light weight system capable of delivering power equivalent to a conventional actuator but is 1/8 smaller in volume. A second version of the AKROD is designed using a brushless DC motor and a planetary gearbox to evaluate the design, torque delivery and other parameters that influence the performance of the brace. Position and impedance control techniques has been developed and implemented on a test bed system and have successfully evaluated and quantified the performance of the actuator to mimic the human gait pattern.
ACKNOWLEDGEMENT

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Thanks to Dr. Paolo Bonato from Spaulding Rehabilitation Hospital for his time and expertise during the entire course of this project. Special thanks to Iahn Cajigas for his guidance with the controls aspect.

Thanks to WGI, for collaborating on this project and helping us fabricate the various parts of the brace.

I also thank the rest of the “Robotics Crew” for making such an exciting environment to work. Richard, Mark, Maciek, Ozzy, Yalgin and Ye – thanks guys for all your support and assistance. It’s been an awesome ride!

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CHAPTER 1 INTRODUCTION

1.1 Motivation

According to the National Stroke Association, stroke is the third leading cause of death in America and a leading cause of adult disability. On the other hand there are more than 6 million people in the United States itself that have survived a stroke [1]. Apart from other diseases that impair patients, the ability to walk is one of the primary pathology that effect post-stroke patients. Abnormalities like muscle weakness, functional deformity, sensory loss and pain impose constraints on the mechanics of walking [2] and these therefore result in either complete loss of their walking capabilities or have deviated walking pattern that could further result in other deformities.

Post-stroke survivors undergo rehabilitation in ways that is labor intensive and physically demanding for the therapist. This means that the patient can receive only a limited amount of exercise time due to shortage in training personal. However in the recent years, robotic technology has been proven effective to overcome this problem as well as it has shown to have advantages over conventional therapy techniques [3]. Treadmill based devices like Lokomat (Hocoma AG, Switzerland) and Autoambulator (HealthSouth USA) have been in clinical use for a few years now. These devices provide active assistance to a) patients who have no motor control by helping them move along predefined fixed trajectories and b) patients who a certain level of walking capability by encouraging them to influence their gait while still providing sufficient guidance.
Although these devices provide optimum level of control to improve patient’s ambulation levels, they are very highly expensive (> $200,000) and occupy large real estate; thereby making it expensive both for the hospitals as well as the patients. Unfortunately the end result is that patients are able to obtain only a limited number of training sessions during rehabilitation. Thus there lies an opportunistic technological gap for a new breed of rehabilitative orthotic devices that has the positive attributes of the treadmill devices while downplaying their faults.

The goal of this project is to develop a wearable and portable smart training orthotic device that could be used by stroke patient throughout daily activities, with constant reinforcement of the targeted gait pattern. The device is called as Active Knee Rehabilitation Orthotic Device (AKROD) as it provides active assistance by generating torque or resisting movement to patient’s knee joint during the entire gait cycle. The AKROD should be comfortable to use, aesthetically pleasing to the patient and can be used on a treadmill and/or over level ground for ambulation purposes. We hypothesize that the constant reinforcement of gait retraining in a real-world environment has the potential to provide more effective and faster gait retraining, thus improving one’s ability to ambulate.

Another motivating factor to make this device light and portable is to enable patients to use AKROD a home environment in conjunction with virtual reality (VR). It has been proven in various studies [4], [5] that the use of virtual reality with robotic rehabilitation system has improved results compared to conventional techniques. For the AKROD, by using the VR game software the therapist can prescribe a desired force-field trajectory for the patient to follow which can be remotely monitored in real-time using the internet.
Overall, by using the combination of AKROD and Virtual Reality we hope to see the following benefits:

a) Lesser resources required by the hospitals (number of therapists, time spent by the therapist with individual patients, etc.); therefore reduced medical cost

b) Patient save time and cost required to visit the hospital

c) Patient exercises and improvements can be monitored regularly by the therapist

d) Increased motivation for the patients to exercise at home even without direct medical supervision.
CHAPTER 2 BACKGROUND

This chapter discusses the medical background involved with the human gait and the knee joint, kinetics and kinematics of the gait cycle, gait pathologies etc. Further various existing methodologies for rehabilitation and current state of the art orthotic devices are also discussed.

2.1 Normal Gait

Normal human walking is defined as ‘a method of locomotion involving the use of the two legs, alternatively to provide both support and propulsion’ [6]. Gait and walking are commonly used interchangeably. However the difference between the two is that, gait describes the manner or style of walking, rather than the walking process itself. Also it is significant to discuss about the difference in the gait between two individuals than about the difference in walking.

2.2 Terminology

Some of the important terminologies relevant to the gait analysis are discussed below [6]:

Anatomical Positions: Six terms are used to describe the directions with relation to the centre of the body. They are anterior, posterior, superior, inferior, left and right.
The motion of the limbs is described using the reference planes.

(a) Sagittal Plane: Divides the body into left and right portions.
(b) Coronal Plane: Divides the body into front and back portions. It’s also known as the frontal plane.
(c) Traverse Plane: Divides the body into upper and lower portions. It’s sometime referred as horizontal plane.

Figure 1: The anatomical position, with three reference planes and six directions
Gait Cycle: It’s defined as the steady-state movement of normal locomotion in a repeating cycle [7]. Generally the beginning and end of the cycle is considered with the ground contact of the same foot. Based on the picture shown in Figure 3, the right foot (reference foot) contacts the ground with the heel and ends when it contacts the ground again.

2.3 Phases of Gait Cycle

The cycle is divided in mainly 2 phases, namely stance phase (time the reference foot is in contact with the ground) and swing phase (time the reference feet is off the ground) [7]. The movement of both the limbs that occur during the gait cycle is known as the stride cycle.
The stance phase of the gait consists of approximately 60% of the gait cycle and rest 40% is the swing phase. At around 10% of the gait cycle, the left leg leaves the ground to begin the swing phase and return back at around 50% of the gait cycle. Thus the gait cycle consist of 2 periods (each lasting around 10%) in which both the limbs are in contact with the ground. These periods are called double limb support, and the remaining cycle is single limb support.

The swing phase is divided into early, middle and late periods. The early swing continues from 60% to 75% of the gait cycle and is distinguished by the rapid withdrawal of the feet from the ground. Mid swing continues until around 85% of the gait cycle and consist
of the period where the swinging limb passes the stance limb. Finally the late swing period brings the limb towards the ground for contact.

![Figure 5: Swinging Phase of the gait cycle consists of early, mid and late swing period](image)

### 2.4 Gait Kinematics

The free rigid body moving in a 3-D space has a total of six degrees of freedom – three linear movements and three rotational movements [8]. At any given time, these movements can be described by in terms of the following variables - displacement, velocity and acceleration. The study of gait analysis in terms of these variables is called kinematics.

The convention used to define the knee angle is as follows [9]:

\[
\text{Knee Angle} = \theta_{21} - \theta_{43}.
\]

If \( \theta_{21} > \theta_{43} \), the knee is flexed and if \( \theta_{21} < \theta_{43} \), the knee is extended.
2.5 Gait Kinetics

Kinetic analysis of the motion of a body involves the relationship between forces and torques required to accelerate the body into motion. The muscle moment calculation is based on Newton’s third law [10]. This means that if the muscle acting at a joint is producing a moment, then there must be other forces acting to produce an opposing moment. One of these forces that are always present is gravity acting at the centre of mass of the body. Although the gravitational force is constant, the moment produced depends on the location of the centre of mass and the centre of rotation. The second force opposing the muscle force is called the inertial force (arises from the inertial properties of the body segment). It is proportional to the acceleration of the segment, but acts in the
opposite direction. The third force is the ground reaction force exerted by the walking surface of the foot. It is the force developed due to the reaction of the forces generated by the moving body.

Figure 7: Knee moment produced due to Gravitational (G), inertial (I) and ground reaction (R) forces

Figure 7 illustrates the moments about the knee due to the three forces acting on the shank/foot. At the time of walking (around 40% of the gait cycle), all three forces are producing a counter clockwise (extension) moment and the knee joint are producing a clockwise (flexion) moment.
2.6 Gait Physiology

2.6.1 Knee

The knee is a junction of the two long bones – femur and tibia. Small arcs of motion result in significant changes in either foot or body locomotion [2]. Therefore knee mobility and stability are major contributors to the normal walking in human beings. During the stance phase, the knee is the basic determinant of limb’s stability, and for the swing phase, knee flexibility is the primary factor for limb’s advancement. Also, the number of two joint muscles involved in knee control indicates its close coordination with the hip and the ankle.

The knee joint is characterized by ranges of motion in the sagittal plane, coronal plane and transverse plane. Sagittal motion (flexion and extension) is used for progression in stance and limb advancement in swing phase. During each stride cycle, the knee joint undergoes four arcs of motion, with alternating flexion and extension as shown in Figure 8 [2].
The average displacement and sequence of knee motion during each stride is as follows:

Table 1: Knee joint angular motion over a gait cycle

<table>
<thead>
<tr>
<th>Gait Cycle</th>
<th>Displacement</th>
</tr>
</thead>
<tbody>
<tr>
<td>0 - 15%</td>
<td>Flexion to 18°</td>
</tr>
<tr>
<td>15 – 40%</td>
<td>Extension to 5°</td>
</tr>
<tr>
<td>40 – 70%</td>
<td>Flexion to 65°</td>
</tr>
<tr>
<td>70 – 97%</td>
<td>Extension to 2°</td>
</tr>
</tbody>
</table>

Motion in the coronal plane facilitates vertical balance over the limb, and transverse rotation accommodates the changes in alignment as the body swing.

During a stance, the knee creates four torque patterns as shown below. The magnitude of the peak torques and timing within the gait cycle varies with individual to individual;
however the general pattern remains the same [11] – extension, flexion, extension and flexion.

![Physiologic Torque](image)

**Figure 9:** Knee Joint Torques; there are four peaks; two extensions torque each followed by a flexion torque of similar magnitudes (4 and 2 BW. LL) [2]

The physiological torque is expressed in terms of body weight x leg length (BW x LL) to eliminate the variations in the subjects. For this project, the reference trajectory data was obtained by testing normal human adults and is provided in Section 6.2

### 2.7 Gait Pathology

#### 2.7.1 Knee Pathology

The abnormalities that prevent natural gait or mechanics of walking are generally categorized as: deformity, muscle weakness, impaired motor control and pain [2]. This document focuses on pathologies related to the knee and issues related to the motion of the knee joint.
The most common types of knee abnormalities occur in the sagittal plane. There are four identified errors for clinical purposes and reinforce functional significance –

a) Inadequate flexion – failure to accomplish the normal range of flexion, resulting in limited or no motion
b) Excessive flexion – more than normal range
c) Inadequate extension – persistent flexion at a time the knee normally extends
d) Excessive extension – motion beyond normal. Consists of extensor thrust and hyperextension.

In the context of the above, it is useful to define the following

a) Extensor Thrust: the effect of excessive extension force when the knee lacks a hyperextension range.
b) Hyperextension: occurs when the knee has the mobility to angulate backwards.

![Diagram of knee functions over a gait cycle]

Figure 10: Abnormal Knee Functions over a gait cycle; IC: initial contact, LR: Loading Response, MSt: Mid Stance, TSt: Terminal Stance, PSw: Pre Swing, ISw: Initial Swing, MSw: Mid Swing, TSw: Terminal Swing
Table 2: Gait Deviations at the Knee [2]

<table>
<thead>
<tr>
<th>Inadequate Flexion</th>
<th>LR</th>
<th>MS</th>
<th>TS</th>
<th>PSw</th>
<th>ISw</th>
<th>MSw</th>
<th>TSw</th>
</tr>
</thead>
<tbody>
<tr>
<td>Excessive Extension</td>
<td></td>
<td></td>
<td></td>
<td>X</td>
<td>X</td>
<td>X</td>
<td></td>
</tr>
<tr>
<td>Extensor Thrust</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>X</td>
</tr>
<tr>
<td>Hyperextension</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>X</td>
</tr>
<tr>
<td>Excessive Flexion</td>
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<td></td>
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<table>
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<th>LR</th>
<th>MS</th>
<th>TS</th>
<th>PSw</th>
<th>ISw</th>
<th>MSw</th>
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<tr>
<td>Valgus</td>
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2.7.2 Hemiplegic Gait

The three main function of human walking are [8]

a) to move from one place to another
b) to move safely and
c) to move efficiently

All three goals are significantly compromised when a patients survives a neurological event. Stroke patients display a gait pattern that varies significantly in some case when compared to normal gait. Although differences occur from patient to patient, some generalities have been demonstrated. Hemiplegic patients have demonstrated differences in temporal, kinetic and kinematic factors and muscle activation patterns across multiple joints.

The following general deviations have been demonstrated [8]:

- Decrease in walking velocity
- Shorter duration of the stance phase for the affected limb
- Decreased weight bearing for the affected limb
- Increased swinging time for the affected limb
- Increased stance time for the unaffected limb
- Decreased step length for the unaffected limb

In most stroke patients, the same joint phases that exist in normal gait are present in both the affected and the unaffected limb; however the range of motion is significantly compromised. Some of the common pathologies that exist with reference to the gait cycle are:

- During the stance phase: abnormal base of support (toes flexed) and limb instability (knee buckling or hypertension)
- During swing phase: inadequate limb clearance (toe drag) and limb advancement (limited hip or knee flexion)

### 2.8 Current Methods of Gait Retraining

In the treatment of knee injuries the overall lower extremity muscle strengthening, improvements in flexibility and technique refinements are more important than the type of the brace itself [12]. However in the following section various methods that are currently being employed by the physical therapist are discussed.

#### 2.8.1 Balance and Strength

Regardless of the gait training model, it is strongly believed that there are at least three requirements for gait [13]. They are

a) balance of head, arms and trunk

b) limb support and

c) limb advancement.
In regards with this, it is proven that the only way to improve locomotor related balance is to practice locomotor type activities. It has been proposed that strength performance not only depends on the quantity and quality of the muscle involved but also upon the ability of the nervous system to appropriately activate the muscles. Hence strengthening involves not only changing the size of the muscle, but also learning to activate the muscles appropriately in the proper sequence, timing and magnitude.

2.8.2 Stretching

Stretching is employed as one of the major techniques usually during the initial stages of rehabilitation with the goal to restore some degree of mobility and muscle control to the affected areas of the limbs. A therapist moves the joints manually in order to stretch the muscles or the patient will perform isokinetic/isotonic exercise using a certain amount of resistance. The biggest drawback to this method is that little or no attention is given to the actual task of improving the gait of the patient. Also exercises require a large amount of therapist’s time and effect, thereby making the process significantly expensive. Another case to be considered is that the effort put in by the therapist will not be consistent over various sessions even when the task is performed by the same person.

Figure 11: Stretching for muscle strengthening
Recently, there has been some work done towards the development of intelligent stretching devices for ankle joints with contracture/spasticity [14]. Although it provides a quantifiable method for analyzing stretching, it does not address the gait rehabilitation directly. More so, no similar device has been developed for the knee specifically.

2.8.3 Brace and Orthoses

According to the American Academy of Orthopedic Surgeons (AAOS), commercially available knee braces are classified into four groups: prophylactic, rehabilitative, functional, and patellofemoral [15].

Prophylactic Knee Brace: are designed to prevent or reduce the severity of knee injuries.

Rehabilitative Knee Brace: are designed to allow protected motion to an injured knee or knees have undergone a surgery during the rehabilitation.

Functional Knee Brace: are designed to provide stability to the Anterior Cruciate Ligament (ACL) and other ligament deficient knee and also provides protection for the ACL and other ligaments after repairs or reconstruction

Patellofemoral: are designed to provide pain relief to the arthritic knee.

Figure 12: (a) Prophylactic (b) Functional (c) Rehabilitation (d) Patellofemoral knee braces [16], [17], [18]
Of these, functional and rehabilitative knee braces have been proven to be effective. On the other hand, their correct application depends on the rehabilitation processes and other activities during the process of rehabilitation.

2.8.4 Treadmill Training

Walking on the treadmill has been suggested as a treatment intervention for both pediatric and adult population with neuromuscular disorder [13]. The use of treadmill could assist the patient by forcing them to speed up their gate. It is largely believed that if the patient is matching the speed of the treadmill then the patient is in turn strengthening the activated muscles. Another advantage of this type of training is to gain the endurance without having to handle the requirements of over ground locomotion i.e. person or obstacle avoidance etc.

Figure 13: Locomat®, robotic system designed to facilitate treadmill gait retraining
The disadvantage of treadmill training are: speed of the walking is being externally driven by the treadmill rather than the patient’s feet, the patient is not pushing off to initiate swing rather lifting up to keep up the motion of the treadmill and the stance limb is pulled backwards under the trunk instead the trunk gliding forward over the stance limb. With these arguments, it has been debated over the effect that the treadmill training will have over the actual over ground locomotion.

With the advancement in the robotics technology in the recent years, treadmill training now is being used in combination with an overhead lift system. The training takes place such that the patient’s weight is either fully or partially supported while the robotic system assists the patient’s locomotion over the belt of the treadmill. One such system called the Lokomat®, Hocoma AG, and Switzerland is shown below.

### 2.9 Current State of Art Orthotic Devices

An orthosis is defined as a device mainly designed to align, correct or prevent neuromuscular or musculoskeletal dysfunction, disease, injury or deformity [19]. Based on patient and therapist interaction, devices are mainly classified into three main categories – passive, active assisted and active resistive [20]. However for the purpose for a better understanding of the devices currently under research and commercially available, they’ve been classified as passive devices, semi active devices and active devices as defined below.

Passive Devices: These use the patients’ body forces to move the leg. It provides stability and help to maintain alignment of joints. Although stability is important part of the
rehabilitation, by using passive devices it is done at the cost of mobility, thus compensating for any gain in muscle strength.

Semi Active Devices: Recently designed orthotic devices employ torsion springs, pistons and simple mechanical components that can be categorized as “semi active”. Innovative designs allow adjustments of certain components such as resistance and damping to provide further flexibility and improving effectiveness over a simple passive devices.

Active Devices: These devices are much complex in design but are more versatile in performance by employing actuators or other alternatives to help/assist the motion of the human limbs.

Based on the above classification of devices, the following section discusses the various devices that were/are under development.

Banala et al. designed a passive gravity balancing passive leg orthosis [21]. The device can partially or fully balance the gravity of the human leg over its range of motion and it is adjustable to the geometry and inertia of a specific patient to achieve the desired level of gravity balancing. Gravity balancing is achieved by first locating the centre of mass of the human limb and the orthosis and then springs are added so that system is gravity balanced in every configuration
Some of the active devices for lower extremity rehabilitation are Powered Leg Orthosis [22], Hybrid Assistive Limb (HAL) [23], Rewalk [24] and Ankle Foot Orthosis [25].

Banala et al. [22] connected the gravity balancing device [21] to a walker and a rigid frame as shown in Figure 15. The trunk had four degree of freedom (dof) with respect to the walker - the hip joint has 2 dof, knee has 1 dof and ankle 1 dof.
Figure 15: Powered Leg Orthosis;

In Figure 15 the following refers to - A: boom to support hip motor, B: hip linear actuator, C: spring-loaded winch to support device weight, D: walker to support the device, E: treadmill F: hip joint, G: load cell on hip linear-actuator, H: knee linear actuator, I: knee joint J: load-cell on knee linear actuator.

Hybrid Assistive Limb (HAL) [23] is an exoskeleton, powered assistive suit designed to aid person with disability for movements such as standing up, sitting down and going up and down the stairs in addition to normal walking. HAL operates by using the EMG signal obtained from the user’s skin via electrodes as the input parameter to determine their intent and then accordingly an appropriate torque is delivered to the joint to produce the movements. Shown in Figure 16 is the full body suit version of the device.
According to the designers of HAL [26], the intended application of this device in the medical field is for rehabilitation support, physical training support, activities of daily living (ADL) support for disabled people. Other applications include support for workers in heavy industry and rescue workers.

ReWalk designed by Argo Medical Technologies, Israel is also a wearable robotic suit [27]. The users initiate the mobility and are able to walk with the assistance of crutches, controlling the suit movement through changes in the centre of gravity and upper body movements.
As mentioned earlier, treadmill based training devices are usually employed in the clinical setting where the patient body weight is supported and the robot assists in the patient’s limb movements. Lokomat is widely used amongst the other treadmill based devices such as Auto Ambulator developed by HealthSouth [28], Mechanized Gait Trainer [29] and LOPES [30]
Figure 18: Autoambulator by Health South

Figure 19: Mechanized Gait Trainer
Magnetorheological (MR) fluid based devices have been designed both for active assistance [21] as well as resistive torques [31], [32]. Chen et al. [33] have designed a magnetorheological (MR) fluid based leg exoskeleton using it with DC motors.
Active torque is provided by the DC motor and the MR actuator as a clutch to transfer the torque to the leg. However when passive torque is required the DC motors is turned off and the MR actuator as a brake to provide resistance.

Electro-rheological Fluid (ERF) based actuator has also been used to design a resistive torque device [34] for the knee joint specifically for isokinetic and isotonic rehabilitation purpose. Along with the design of ERF components, the knee brace was also designed and fabricated. The target market for this device was low cost, portable and compact orthosis.

Figure 22: ERF based Knee Rehabilitation Device using variable controlled resistance. Also shown is the ankle and waist support to assist the knee device

Another set of active devices not directly designed for the application of rehabilitation but are similar in terms of functionality are RoboKnee [35] and BLEEX [36]
RoboKnee is a one degree of freedom exoskeleton device designed to offset the energy required by the users to work against gravity. It uses low impedance series elastic actuator between the upper and lower potions of the knee brace so that it provides torque about the knee joint as shown in Figure 23.

![Figure 23: RoboKnee](image)

BLEEX, Berkeley Lower Extremity exoskeleton is a power assisted device designed for human strength and endurance environment by adding extra force to the lower limb movements.
Another set of passive device is the ones developed by Honda [37] and has been in the news recently. It is an assistive walking device with bodyweight support system to reduce the load on leg muscles and joints such as the hip, knees, and ankles.
2.10 Conclusion

From the previous section it is seen that a variety of devices with the different technologies and concepts have been developed for rehabilitation purposes. Each of these devices and technologies has their advantages and disadvantages. For example, the treadmill training devices provides actuation for various joint (hip, knee and ankle) and provides a system for body weight support. On the other hand, it is extremely expensive and occupies large real estate thereby making it expensive for the hospital and the patients. Passive devices are simple in terms of technology but cannot be adjusted in real-time or provide the versatility for the patient requirements.

Therefore it is concluded that their lies a opportunity in terms of technology as well as from a commercial point of view to develop a new breed of low cost rehabilitative orthotic device that can actuate the knee joint to provide reinforcement for post-stroke patient that have recovered a certain level of ambulation capability but have not completely attained their normal walking gait pattern. The challenge here was to be able to have the advantages of the treadmill training devices but at the same time the device be reasonably light weight, compact and other attributes so that the patients will be able to comfortably use it an home environment setting.


CHAPTER 3 DESIGN OVERVIEW

3.1 Design Requirements

This chapter explains the targeted specification and requirements of the Active Knee Rehabilitation Orthotic Device (AKROD). The first and foremost goal of AKROD was that the device be safe to use at all times, both for the patient and the therapist involved. With this is mind the following design goals were summarized:

- The device be able to provide active assistance to the patient in the advancement of his/her gait cycle
- Device be able to generate the required torque for ambulation
- The device weigh less than 3.5 lbs (1.6 kg); 50% reduction from the previous version [34]
- The system controller be able to control the joint position and moment during the entire gait cycle
- The actuator be placed as close to the knee joint
- Alternate location for actuator placement is closer to the centre of mass of the human body. This would reduce the weight perception felt by the patient mainly during the swing phase of the gait cycle
- Mechanical stop to prevent hyperextension
- Ability to use over flat ground, on a treadmill with slight inclines and declines.
- Attachment to the patient be possible while he/she is seated on the flat bench
• No hardware on or around the patient’s leg that will impede their overall ability to ambulate with the device on them

• Comfortable for use, both for the patient and the therapist

Overall, the AKROD was designed to assist stroke patients during the post-stroke rehabilitation stage that will enable them to regain their normal walking capability. Therefore the addition of an exoskeleton to their body should facilitate their rehabilitation process rather than impeding it.

The target parameters for the above mentioned design goals are discussed in detail below.

• **Range of Motion**
  
  Based upon the normal human walking data at normal speed, along the sagittal plane the flexion-extension knee joint angle ranges between -5° and 60° (flexion). Therefore it was required that the mechanical stops be included to prevent the maximum extension (also known as the hyperextension) and also be adjustable to meet individual patient needs. The range of adjustment required was between -5° to 0°.

• **Power Output and Speed**
  
  According to [2], [9] and reference trajectories obtained from the tests conducted with healthy subjects at Motion Analysis Lab, Spaulding Rehabilitation Hospital (See Section 6.2), for a healthy human at a normal walking speed (approximately 1.32m/s) a maximum peak moment of about 54 Nm is generated by the knee joint during the stance phase (at approx. 10% of the gait cycle). The peak knee velocity of about 7
rad/sec occurs at approximately 90% into the gait cycle. The knee kinetics occurring at 10% and 90% of the gait cycle is shown in Table 3.

Table 3: Peak Kinetic values occurring during the gait cycle¹

<table>
<thead>
<tr>
<th></th>
<th>10% of GAIT CYCLE</th>
<th>90% of the GAIT CYCLE</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Moment</strong></td>
<td>54 Nm (max)</td>
<td>27 Nm</td>
</tr>
<tr>
<td><strong>Velocity</strong></td>
<td>2 rad/sec ≈ 19 rpm</td>
<td>7 rad/sec ≈ 67 rpm (max)</td>
</tr>
<tr>
<td><strong>Power</strong></td>
<td>108 W</td>
<td>189 W</td>
</tr>
</tbody>
</table>

From a design standpoint, the device had to provide the patient an assistance level ranging between 25% - 50%. In other words, at 25% assistance level, the actuator would have to provide at least 25% of torque and the patient would provide the remaining 75% torque in order to advance their gait. And at 50% assistance level, 50% of the torque is provided by the actuator and the remaining 50% is provided by the patient. Using this design goal and knee joint values, the following moment and power values were calculated (shown in Table 4):

Table 4: Targeted kinetic parameters

<table>
<thead>
<tr>
<th></th>
<th>10% of Gait Cycle</th>
<th>90% of Gait Cycle</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>50% assistance</strong></td>
<td>Moment</td>
<td>27 Nm</td>
</tr>
<tr>
<td><strong>25% assistance</strong></td>
<td>Power</td>
<td>54 W</td>
</tr>
</tbody>
</table>

Based on above parameter, the actuator will have to provide power of about 95W (peak) in order to provide 50% assistance. However taking the size of the actuator,

¹Reference body weight of 135kg was used to determine the torque values
placement and weight issues into consideration, the actuator should be able to
generate no less than 50W power (peak). At 50W, the device will meet the 25%
active assistance goal.

- **Actuator Placement**
  The placement of the actuator in reference to the patient’s body was a critical design
consideration. There were a couple of factor that needed to be taken into
consideration.

  **Inertial Effect:** Any additional mass added to the person’s body will create an
inertial effect which can be detrimental to the rehabilitation process. From a
mechanical standpoint the ideal location for placement of the actuator is at the knee
joint. By placing it at the knee joint, the power transferred in terms of distance is at
minimum and reduces design complications. However on the other hand, due to
weight of the actuator there will inherent inertial effect felt by the patients especially
during the swing phase of the gait cycle. In order to overcome this problem to a
certain extent the actuator should be placed closer to the body’s center of mass. The
closer the actuator is to the centre of mass, the lesser physiological effect felt by the
patient.

  **Twisting Effect:** From past experience [34], it was learnt that when the knee joint is
actuated only on one side, it creates a twisting effect on the brace, thereby causing the
brace to migrate from its actual location. Therefore it was concluded that power (or
torque) had to be transferred to both sides of the knee joint (medial and lateral) to minimize the twisting effect.

- **Disengagement**

  It was required that actuator be able to disengage itself such that no power is transferred to the knee joint even when the device is attached to the patient’s body for the following two reasons:
  - Therapist can obtain baseline measurements to determine the level of assistance that the patient might require
  - In case of emergency, if the actuator fails the brace position (and the patient’s leg) would have to be quickly adjusted.

- **Mechanical Hyperextension Stops**

  With one of the main goals of the device being able to prevent hyperextension, it was required that the design include mechanical stops such that the patient does not hyperextend beyond a certain limit. It was also required that these mechanical stops be adjustable between -5° to 0° to meet the patient’s individual needs, rehabilitation stage and gait capabilities.

- **Hinge Type**

  The nature of biomechanics of the knee joint is such that the femur slides backward and forward along the tibia plateau as it flexes and extents. Due to this, the centre of
rotation is constantly changing. This is called a polycentric rotation. For a mechanical device to be able to follow this motion, a complex polycentric hinge would be required. Upon consultation with therapist and experts in the field it was recommended that the brace have a polycentric hinge. However due to the complexity of such a hinge, alternative suggested solution was to use a “pseudocentric” hinge. In the pseudocentric hinge, the centre line of the upper arm of the brace is anterior to the centre of rotation and the lower arm posterior. This results in a unique motion that is closest to the polycentric hinge.

- **Migration**

Test from previous version of the brace [34] also showed that it had a tendency to migrate and slip down from the attached location. Hence it was important that the brace be designed such that it would overcome this problem either by certain inherent properties that will prevent migration or by using other external solution that can be retrofitted to the brace design.

It was previously found that the use of support at the hip and ankle level can help the brace arm stay in its required position. For the waist support, an off-the-shelf hip brace (Model # 3457) manufactured by Newport was tested. And at the ankle level, an ankle brace manufactured by Donjoy was used. The combination of hip and ankle was shown to have a significant effect on migration. It helped the brace arm maintain its alignment with respect to the centre of the knee joint. Therefore the design of AKROD had to include the hip and ankle attachments.
• **Patient – Device Attachment**

Attachment of the device to the patient’s leg/body was a critical design parameter. Not only does the device have to be safe and comfortable to use, it also had to be rigidly attached to their body such it moves along consistently along with the patient’s leg movement to ensure minimized controller discrepancies and improved robustness. The device had to be compatible to a varied range of population with little or almost no modification. And also the patient or the therapist should be able to remove the device easily and comfortably both in normal situation as well as in case of emergencies.

![Figure 26: Cuffs for Patient-AKROD Interface](image)

One of the solutions for patient attachment that was highly recommended by the medical personal at Spaulding Rehabilitation Hospital was “cuffs” as shown in Figure 26. These cuffs are currently being used on Lokomat. It snugly wraps around the patient’s thigh and shanks to attach the device to the patient’s body. They are
extremely comfortable and easy to use, both from the patient’s end as well as for the therapist.

The added advantage of cuffs is that they also help overcome the migration issues. Also since the cuffs are available in various sizes, it will allow a universally sized brace to be used for varying patient body dimensions.
3.2 Actuator Selection

Amongst other performance characteristics, the most important of them was power density of the motor (torque to weight ratio). AKROD being an orthotic device that would be attached to the patient’s body, any extra mass had to be carefully considered. With this in mind, the choice was limited to rotary actuators. Other actuators like pneumatic, hydraulic and linear electromagnetic were considered but were deemed unsuitable for this application either due to weight requirements or other performance characteristics. Further research was done into various types of DC motors which are discussed below.

3.2.1 Brushed DC Motor

Brushed DC motor basically consists of a stator, rotor, brushes and a commutator. The stator generates a stationary magnetic field (using either permanent magnets or electromagnetic windings) that surrounds the rotor. The rotor, also known as the armature, is made up of one or more windings that are connected to a mechanical commutator. The magnetic field developed by the rotor is attracted to the opposite poles of the stator, causing the rotor to turn. As the rotor turns, the windings need to be constantly be energized in different sequence. This requires a change in the direction of the windings and is done by brushes and commutators. The brushes are attached to the motor’s external wires and the commutator segments slide over the brushes so that the current switches at the correct angle.
3.2.2 Brushless DC Motor

Similar to brushed motors, brushless DC motor works on the same principle of magnetic field switching. However brushless DC motors are electronically commutated rather than using brushes. In the servo motor version, this is accomplished using Hall Effect sensors or encoder feedback. They are mainly designed for high performance applications where high speed and continuous torque are essential requirements. Also brushless DC motor allows for better heat dissipation because the windings are located on the stator.

Shown below in Table 5 details some of the advantages and disadvantages of brushed and brushless DC motor.
Table 5: Brushed DC Motor vs. Brushless DC Motor

<table>
<thead>
<tr>
<th></th>
<th>Advantages</th>
<th>Disadvantages</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Brushed DC Motor</strong></td>
<td>easy to control due to the mechanical commutation of the brushes</td>
<td>tend to wear out quickly in high torque applications due to high currents drawn</td>
</tr>
<tr>
<td></td>
<td>easy to decrease cogging by increasing the number of slots</td>
<td>tend to give off sparks at the brush interface</td>
</tr>
<tr>
<td></td>
<td>only an analog output and an amplified are needed</td>
<td></td>
</tr>
<tr>
<td><strong>Brushless DC Motor</strong></td>
<td>designed high precision application</td>
<td>increased controller complexity</td>
</tr>
<tr>
<td></td>
<td>can be operated at high torque for a longer period of time</td>
<td>difficulty in reduced cogging</td>
</tr>
<tr>
<td></td>
<td>can be lower speeds for a longer period of time</td>
<td></td>
</tr>
<tr>
<td></td>
<td>cogging can be compensated with a fast controller</td>
<td></td>
</tr>
</tbody>
</table>

Based on the above discussion, brushless DC motor was considered the best option for AKROD mainly due to better performance during high torque, long speed operation as well other factors like better heat dissipation and long life. Also it is important to note that due to high torque and low speed required for this application, a gear reduction would be required in order to reduce the overall weight of the actuator. Therefore any chosen motor would need a compatible gear reduction component.

3.2.3 Gear Bearing Drive (GBD)

Another actuator technology that was considered was called Gear Bearing Drive (GBD). Gear Bearing Drive (patent pending) is a compact actuator technology that provides two solutions – actuation and joint providing support. It consists of a combination of a
brushless DC motor (Outrunner type) and Gear Bearing technology (NASA developed and patented). Further information and discussion on Gear Bearing Drive is provided in the Section 3.3.
3.3 Gear Bearing Drive (GBD)

3.3.1 Overview

Gear Bearing Drive (patent pending, Application # 11/821,095) is a novel technology that combines a brushless DC motor and gear bearing into a single mechanism. The resulting compact device provides two solutions: a) it operates as an actuator and b) it provides a joint support. The Gear Bearing Drive replaces the traditional motor gear train assemblies thereby reducing weight and space.

The principle motor component of the GBD is a standard off-the-shelf low cost brushless DC “outrunner” motor most commonly used on hobby airplanes and racing cars. The outrunner motor is 3-phase, 12 pole brushless DC motor with an inner stator and an outer rotor. The magnetic coils are mounted to the end bell (grounded stator) and the permanent magnets are attached to the rotating can (rotor). These motors are ultra-compact with diameters between 1 to 2 inch and lengths ranging between 0.87 and 2.63 inch depending on the power output required. They operate at high speed of 8000 – 15000 rpm.

The other main component of the GBD is the gear bearing system that was developed and patented by NASA. The gear bearing is a novel bearing less high-reduction ratio planetary gear system which places a rolling surface at the pitch diameter of each gear to maintain the gearset alignment to support thrust, radial and bending loads.
In the Gearing Bearing Drive (GBD), the torque transfer from the actuator occurs through two stages of planetary gear system, namely the input stage and the output stage. The core principle of the gear bearing exists with the difference in the number of teeth on the planet gear on the input stage and the output stage.

As shown in Figure 30, the DC motor’s outrunner is press fit into the sun gear component of the input stage. The planet gear teeth, both input stage and the output stage are crowned on the same part. However the number of teeth crowned on the input is “x” and the number of teeth on the output side is “x+1”.

Figure 28: Hacker 3-phase DC "outrunner" style brushless motor

Figure 29: 1.25" diameter gear bearing prototype with a 325:1 reduction ratio
The input stage planetary system (motor, planets and ring) and the planets on the output stage are fixed such that they revolve around the same central axis driven by the motor (sun gear on the input stage). When the motor is turned on, the outrunner moves the planetary gear and the rollers on the input side at the same speed. This causes the planetary gear teeth on the output stage to rotate at the same speed as the input stage. Since the planetary gear on the output stage has one more teeth than the input stage and both have to move at the same speed, this forces the ring gear on the output stage to slowly rotate in order to compensate. The roller on both the input and the output stage provides thrust and roller support thereby eliminating the need for traditional ball bearings.
The novelty of the GBD lies in the fact that by varying the combination of the number of teeth on the planetary gear system can result in a speed reduction ratio that can range between 20:1 to 2000:1 without altering the overall form factor of the device.

### 3.3.2 GBD vs. Conventional DC Motors

A preliminary prototyped of the GBD was design and fabricated using a CAD Rapid Prototyping (RP) machine, off-the-shelf DC motor and controller for comparing it to a conventional type DC motor. A Bodline Electric Gearmotor was used as a reference. Figure 31 and Table 6 below shows the comparison of the two types of actuators for performance, power to weight density and form factor.

![Figure 31: GBD vs. Bodline Gearmotor](image-url)
### Table 6: Specifications Comparison: GBD vs. Bodline Gearmotor

<table>
<thead>
<tr>
<th></th>
<th>Gear Bearing Drive</th>
<th>Bodeline Gearmotor</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Dimensions</strong></td>
<td>Diameter: 76.2 mm (3 in)</td>
<td>Diameter: 90mm (3.55 in)</td>
</tr>
<tr>
<td></td>
<td>Width: 41.2 mm (1.42 in)</td>
<td>Length: 210 mm (8.26 in)</td>
</tr>
<tr>
<td><strong>Volume</strong></td>
<td>187 cm$^3$ (10 in$^3$)</td>
<td>1336 cm$^3$ (81.7 in$^3$)</td>
</tr>
<tr>
<td><strong>RMP</strong></td>
<td>266</td>
<td>263</td>
</tr>
<tr>
<td><strong>Torque</strong></td>
<td>3.42 Nm (30.3 in-lb)</td>
<td>2.37 Nm (21 in-lb)</td>
</tr>
<tr>
<td><strong>Gear Ratio</strong></td>
<td>40:1</td>
<td>9.4:1</td>
</tr>
</tbody>
</table>

#### 3.3.3 Conclusion

It can be seen from the above discussion that the Gear Bearing Drive technology is superior and highly adaptable for the robotic application. The GBD can combine an actuator and joint load supports into a space that is volumetrically smaller than existing actuating technology for a similar power density output. Henceforth it was decided that the first prototype of the AKROD would be designed and fabricated to evaluate the GBD as an actuating mechanics.
3.4 Sensor Requirements

Sensors are required to provide kinetic and kinematic data information that can be relayed for measurement purposes in order to be able use it in the controller algorithm. The number and the types of sensors depend on the data that will be required for the control implementation.

In general, position feedback is required for a position controller and force feedback for force/torque controller. Combination of the two can be used for a hybrid controller. For AKROD’s application, it was determined that the following data/information is critical:

- Instantaneous position and velocity of the knee joint angle during any given stage of the gait cycle.
- Instantaneous Torque/force at the knee joint during any given stage of the gait cycle.
- Stage of the gait cycle.

Some of the parameters for sensor selection should include:

- Weight of the sensor: Since the sensor system is attached to the patient’s lower extremity, any additional weight should have a minimum effect on their rehabilitation capabilities.
- Accuracy: The sensor has to provide accurate measurements. This is most critical in determining the knee joint angle during hyperextension.
- Repeatability: The sensor has to provide consistent measurement for the data acquired every time in order to control algorithm to exhibit robustness.
• Range: The position sensor has to have range of motion of the knee joint. The force/torque sensor should be capable of measuring the maximum torque at the knee joint.

• Signal Processing: Most sensor or transducers will require signal amplification and signal condition (noise filtering).

• A/D conversion: Almost all analog sensors will require an A/D conversion depending upon the type of microcontroller being used.

3.4.1 Position Sensor

A position sensor is required to determine the angle of the knee joint. Since the device is actuated along a single axis in the sagittal plane, a single position sensor can be placed on the mechanical system such that it directly measures the knee joint angle flexion and extension. Other locations of the position sensors are possible, but this will increase the complexity of the controller as well as add possible error scenarios. Position sensor information can be mathematically manipulated to obtain velocity.

The most commonly used position sensors are incremental encoders and absolute encoders. Incremental encoder produces square wave pulses as the shaft rotates to measure the relative position of the shaft and starts from a zero count every time it’s turned on. The encoder’s resolution can be increased by using an interpolation factor and additional resolution can be obtained using quadrature decoding.

Absolute encoder overcomes the issue of indexing that is encountered by the incremental encoder. It has a binary disk uses encoding technique which the shaft mounted on the encoder know its exact position, hence the name absolute encoder.
The choice of incremental or absolute is concurrently driven by the mechanical design and low level controller implementation criteria that is discussed in detail in the chapters to follow.

### 3.4.2 Force/Torque Sensor

There are several methods of measuring force or torque. The goal here is to measure the torque at the knee joint during the gait cycle. In the overall system of exoskeleton-patient interaction, there are two torque components – a) torque generated by the actuator and b) torque generated by the patient’s muscles.

Force/torque generated by the actuator and transferred to the knee joint can be locally sensed used force transducers such as load cells. Loads cells are simple in design, long lasting, reliable and easy to implement in the control architecture, hence they were considered for the designs brainstormed.

Torque generated by the patient is a more complex task. This torque is essentially the patient’s intention to either oppose the motion of the exoskeleton or assist the motion or neither of those. It is also related to stage of the gait cycle (stance vs. swing). A number of techniques have been investigated and employed. Some of the methods used are by measuring the ground reaction force [35], using EMG signals [39] and using inverse modeling estimation [40].
CHAPTER 4 GEAR BEARING DRIVE (GBD) 
BASED AKROD 

4.1 Overview 

This chapter discusses the mechanical design aspects of the Active Knee Rehabilitation Orthotic Device (AKROD) using the Gear Bearing Drive (GBD) as the actuating component. The selection of sensors for this design and other components are also discussed in detail. 

This version of the AKROD was designed and fabricated as a proof of concept for two main components: 

- Use of Gear Bearing Drive (GBD) as the actuating component for active assistance to the knee joint 
- Test the brace design in conjunction with the actuator, sensor and other migrating solution that was previously discussed. 

Since this is a prototype version of the device, it was expected that we encounter unforeseen issues which were not realized during the brainstorming and designing stage. It was also prioritized to meet the most important design requirements that would be essential to the overall GBD design and its interaction with the brace.
4.2 Design Specification

Based on the human gait requirements as discussed in Section 3.1, the final prototype of the device using the Gear Bearing Drive has the following specifications:

Table 7: Specification of GBD based AKROD

<table>
<thead>
<tr>
<th>Specifications</th>
<th>Values</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Brace</strong></td>
<td></td>
</tr>
<tr>
<td>% of robot assistance</td>
<td>50%</td>
</tr>
<tr>
<td>Peak Torque</td>
<td>34 Nm</td>
</tr>
<tr>
<td>Range of Motion</td>
<td>0° to 90°</td>
</tr>
<tr>
<td>Total Weight (including GBD)</td>
<td>2.72 kg (6lb)</td>
</tr>
<tr>
<td><strong>Brushless DC “Outrunner” Motor</strong></td>
<td></td>
</tr>
<tr>
<td>Manufacturer</td>
<td>Hyperion Z2213-24</td>
</tr>
<tr>
<td>Motor Type</td>
<td>Outrunner</td>
</tr>
<tr>
<td>RPM/V</td>
<td>850</td>
</tr>
<tr>
<td>Operating Current</td>
<td>6-12 A</td>
</tr>
<tr>
<td>Peak Current</td>
<td>16 A</td>
</tr>
<tr>
<td>Weight</td>
<td>53 g</td>
</tr>
<tr>
<td>Shaft Diameter</td>
<td>22 mm</td>
</tr>
<tr>
<td>Outside Diameter</td>
<td>28 mm</td>
</tr>
<tr>
<td>Length</td>
<td>42 mm</td>
</tr>
<tr>
<td><strong>Gear Bearing Drive (GBD)</strong></td>
<td></td>
</tr>
<tr>
<td>Gear Ratio</td>
<td>167:1</td>
</tr>
<tr>
<td>Material</td>
<td>SLA -40 Plastic</td>
</tr>
<tr>
<td>Outer Diameter</td>
<td>60.35 mm</td>
</tr>
<tr>
<td>Outer Width</td>
<td>97 mm</td>
</tr>
<tr>
<td>Weight(^2)</td>
<td>490 g</td>
</tr>
</tbody>
</table>

\(^2\) Includes the weight of the brushless DC outrunner motor
4.3 Assembly

Due to the complexity of the design, it will helpful to explain the overall assembly first and then followed by the descriptions of each of the individual components in the sections to follow. The final design of the GBD based AKROD assembly is as shown in Figure 32.

The Gear Bearing Drive (GBD) is mounted in front of the patient’s thigh and above the patella. It is centered along their mid section of leg, thus placing it closer to the body’s center of mass. This will minimize the inertia felt by the patient due to the inherent weight of the GBD.

The GBD interfaces with the top end of the push bar rod along its center line and is held in place by fastening it to the lateral and the medial brace. The opposite (lower) end of the push rod interfaces with a U-shaped yolk. The load cell is placed between the push rod and the yolk. The prongs of the yolk are connected to the lateral and medial side of the brace at the knee joint location. When the GBD is turned on, the mechanical power is transmitted to the top end of the push rod by means of an extension shaft. This causes the lower end of the push rod to apply downward force on to the load cell. The load cell then moves the yolk, thereby forcing the lower end of the medial and lateral braces to rotate about the knee joint creating a flexion extension motion.

All the components (otherwise mentioned) were machined from aircraft grade 7075 Aluminum. Its excellent physical properties of higher tensile strength (>500 MPa) and lightweight (density 2.81 g/cm) made it an ideal choice for material selection.
4.4 Gear Bearing Drive Design

The gear bearing drive was modified from the previous prototype shown in Section 3.3 to accommodate the requirements for this particular application.

Using a 167:1, gear ratio, the individual components of the GBD were fabricated using the SLA 40 material on rapid prototyping (RP) machine. The GBD interfaces with the lateral and medial brace using spacers which were also fabricated on the RP machine. The output stage of the GBD has an extension component that interfaces with the top end
of the push bar assembly. The extension component and top end of the push bar are fastened using pin connections.

![Model of Gear Bearing Drive (GBD) actuator used for the Active Knee Brace Orthotic Device (AKROD)](image)

**Figure 33: Model of Gear Bearing Drive (GBD) actuator used for the Active Knee Brace Orthotic Device (AKROD)**

As shown in Figure 34 and Figure 35, the outrunner motor is press fit inside the sun carrier of the input stage of the GBD. Around the sun carrier, there are three sets of pinion carriers separated by angle of 120°. Each pinion carrier has a pinion gear and a pinion roller. The pinion roller rotates around the sun roller and likewise the pinion gear around the sun gears. For simplicity, only one pinion roller and one pinion gear have been shown in Figure 34 and Figure 35. Using the principle of the Gear Bearing design, the input stage pinion gears has 20 teeth (n teeth) and the pinion gears on the output stage have 21 teeth (n+1).
Figure 34: Interior View of the Gear Bearing Drive

Figure 35: Detailed View of the gears inside the Gear Bearing Drive (GBD)
Therefore in order to maintain the speed due to the addition tooth on the output stage, the ring gear on the output stage rotates. Hence making the extension shaft on the output stage rotates along with it. The extension then pushes on to the top end of the push bar assembly thereby transmitting the power to the lower end of the assembly.

The overall weight of the GBD using the SLA-40 plastic is 490 g. It measures 97mm in diameter and 60.35mm in width. The gears are lubricated using lithium grease and the device is sealed off using custom seals.

4.5 Brushless DC “Outrunner” Motor

The brushless outrunner motor used for the actuating the gear in the GBD is model Z2213-24 manufactured by Hyperion. It is an off-the shelf component that is mainly used for hobby airplanes and cars. They are controlled using an electronic speed controller, transmitter and wireless receivers.
Motor installation and control diagram is shown in Appendix A.

4.6 Brace Components

4.6.1 Push Bar Rod Assembly

The push bar rod assembly is the main transmitting element between the GBD and the brace as shown in Figure 37. On the top end it is connected to the GBD’s outer stage extension. The push bar is hollow and is designed to provide the disengagement mechanism which was an essential design requirement. The hollow lower end of the push bar is press fitted with a linear bearing such that an inner connecting rod can freely slide through it.

The inner connecting rod works such that it’s upper end slides into the linear bearing and locked in place using locking pin and the lower end screws into the load cell. When the locking pin is removed, there is no transmission of power from the GBD to the lower assembly, thus allowing the inner connecting rod slides along linear bearing.
In other words, this allows patient can flex or extend their knee joint freely without receiving any mechanical assistance from the GBD.

4.6.2 Load Cell

The mechanical aspect of the load cell interface in the assembly is discussed here.

Due to the nature of the application where the knee shows flexion and extension, the overall force measured by the load cell will see either compression or tension depending on the direction of power transmitted by the GBD and the patient intention (either to oppose or assist the actuator). Hence it was decided that the load cell should have the capability to work in both directions.
The load cell has threaded bars on either ends. In the assembly design, lower end of the inner hollow tube screws onto one side of the load cell and the yolk screws on to the other side. Any axial force (compression or tension) in the assembly will be directly recorded by the load cell. Therefore, using inverse dynamics, the torque at the knee joint generated by the actuator is calculated.

4.6.3 Yoke

Next in the assembly chain below the load cell, is the U-shaped yoke. The load cell is directly screwed into the top end of the yoke as shown. It is symmetrical about the axis passing through the load cell therefore allowing for equal torque to be transmitted to either side of the knee joint. The prongs are mated with the lower brace and allows for a comfortable flexion-extension of the knee joint.
4.6.4 Lateral Brace Assembly

The lateral brace assembly is made up of two frame components – one superior (above) to the knee joint and the other inferior (below) the knee joint as shown in Figure 41 and Figure 42. The frames are aligned using bearings to allow for free and smooth motion for the patient’s knee joint. The resulting rotation creates a pseudo centric joint in which the upper frame is offset from the lower frame. In order to reduce material and overall weight of the frame, triangular cut outs are strategically included in the design without compromising with the strength capability for various loading situations. In addition, the brace thickness is 0.002” which is comparable to the off-the-shelf braces manufactured by companies such as Donjoy etc.

The other components in the lateral brace include potentiometer, adapter plate and bearing retainer components.
The upper frame of the later brace has through holes at the top to allow for attachment of Newport 4 Hip Orthosis. The hip orthosis can be easily added or removed using standard socket head screws. Standoffs are provided for GBD mounting using spacers and also for cuff attachment as shown.
Figure 41: Lateral Upper Brace Assembly

Figure 42: Lateral Lower Brace Assembly
The lower lateral brace interfaces with the upper brace by means of bearings. In order to provide a smooth and free motion of the patient’s knee joint, it was essential that the upper and lower brace be a smooth rotation relative to one another. This was accomplished by incorporating a bearing at the interface of the two braces. A stainless steel ball style bearing was chosen for its design simplicity as well as its ability to withstand nominal loading. The diameter of the bearing was the driving factor of the dimensions of the brace at the knee joint. The bearing slip fit into the brace and is locked in place using retainer rings. The bearing retainer is fastened to the upper brace using flathead screws. The bearing assembly will maintain alignment and spacing within the knee joint to avoid surface contacts of the independently moving brace parts, thus providing a smooth, frictionless motion to the patient wearing the brace.

Figure 43: Bear Assembly and interface with the brace at the knee joint
At the farther end of the lower brace, cut-outs have been provided for attachment to the ankle brace. Velcro straps from the ankle brace can be fed around the holes between the medial and the ankle brace to provide a secure attachment.

### 4.6.5 Medial Brace Assembly

The medial brace assembly of the AKROD is as shown in Figure 44.

![Figure 44: Medial Brace Assembly](image)

The medial brace assembly is the same as lateral brace assembly in terms of bearing interface between the upper and the lower brace, yoke attachment and the attachment for the ankle brace. However there are a few features that differ from the lateral side that is discussed in detail below.
The upper medial brace is slightly shortened and design adjusted for human anatomy on the medial side. One end of the GBD is mounted using spacers such that it is at the center of the upper leg segment. The spacers are fastened to the upper medial brace using screws such that it can withstand the torsional load applied by the GBD. It allows for quick mounting and dismounting of the GBD to the brace.

4.6.6 Cuffs

As discussed earlier, cuffs were an integral part of the design to interface the brace with the human leg segments. In addition to interfacing, the cuffs have been proven to prevent migration and also provide rigidity between the lateral and medial braces. The rigid metal bar was one of the other driving factors in the dimensioning of the brace width.

![Figure 45: Cuffs to interface between the AKROD and the patient](image)
The two ends of the bar are fastened to the braces at the standouts. It is important to note that the cuff that attach to the upper leg segment is wider than the lower leg segment. Using the width of the rigid bar on the cuffs, the yolk width and GBD spacers were designed in order to maintain alignment along the length of the brace. If the need arises to use wider/bigger cuffs, the only component that needs to be replaced are the yolk, rigid bar on the cuffs and the GBD spacer.

4.7 Sensor Components

4.7.1 Load Cell

The load cell chosen for this assembly is model LCFA-500 manufactured by Omega Engineering, Inc. It is a full Wheatstone bridge circuit design miniature load cell. These types of load cells are accurate, repeatable, simple in design and easy to implement. The maximum range of the load cell is 500 lbs which far exceeds the requirements of the maximum torque generated by the actuator or torque experienced at the patient’s knee joint.

Figure 46: Miniature Load Cell
4.7.2 Potentiometer

A rotary potentiometer is incorporated into the design to measure the knee joint angle directly. The potentiometer is placed at the interface of the upper and lower brace on the lateral side. An outer cover plate mounts concentric to the centre of the knee joint connecting the upper and the lower braces. The potentiometer is then fed to the centre of the cover and is prevented from rotating by using a locating lug.

For this design an off-the-shelf rotary potentiometer (Model 140) manufactured by Vishay Intertechnology, Inc. was chosen by its design simplicity, cost and ease of implementation.

![Figure 47: Rotary Potentiometer to measure the instantaneous knee joint angle](image)

4.8 Final Prototype

Shown in Figure 48, Figure 49 and Figure 50 are pictures of the final prototype of the Gear Bearing Drive (GBD) based Active Knee Rehabilitation Device (AKROD). All the metal parts were fabricated using CNC machine at Northeastern University and the GBD gears were made using the rapid prototyping machine in the Biomedical Mechatronics Laboratory, Northeastern University.
The weight of the final prototype was less than 6 lbs (excluding the hip and the ankle brace).

Figure 48: Partial View of the final prototype of the GBD based AKROD
Figure 49: Final Prototype of the GBD based AKROD on a human leg

Figure 50: Final Prototype of the Gear Bearing Drive (GBD)
4.9 Conclusions

Upon the fabrication of the prototype of the AKROD using the GBD, the following conclusions and action that were needed are listed below.

- The gears were fabricated using SLA plastic material for the proof of concept purposes. It will not be able to handle excessive and cyclic loading.
  - For future testing, evaluation and use of the device, the gear of the GBD would have to be manufactured out of metal which can withstand higher loading cycles and yet meet the weight constraints.

- The existing brushless outrunner motor used in the GBD is sensor less. In other words, it uses back e.m.f. to determine its velocity. Therefore when the motor is turned on, it ramps up to a very high velocity (5000 – 6000 rpm) before attaining the desired speed. This is not ideal for AKROD application.
  - The existing outrunner motor has to be retrofitted with Hall Effect sensors to make it a “Sensored brushless outrunner motor” and has to be under servo motor control. Upon successfully sensing the outrunner, it will be able to power on and start a low rpm ranges (<500 rpm). Thus allowing for smaller change in knee joint angle.

- No mechanical stops were included in the GBD based AKROD
  - Mechanical stops have to be included in the design before any testing/evaluation on a human leg can be performed to prevent hyperextension. The location of the mechanical stops has to be adjustable between (0° and -5°) by the therapist to accommodate different patient types.
• Pin and locking mechanics used at the following locations led to mechanical play in the system: a) interface of the yoke and the brace and b) interface of the GBD extension and the top end of the hollow push bar.
  ➢ The mechanical play has to be reduced/eliminated in order to translate the power from the push rod assembly to the knee joint with maximum efficiency

• Pinned connection used for disengagement purposed on the push rod assembly is time consuming for removal especially during an emergency
  ➢ Disengagement has to be less time consuming and relatively easy process.
CHAPTER 5 BRUSHLESS DC MOTOR BASED AKROD (AKROD)

5.1 Overview

This chapter discusses the final assembly of the AKROD using Brushless DC Motor and changes made to previous brace design components. The mechanical components that have not been changed or altered will not be revisited in this chapter. Also discussed are the final product specifications of the device.

After successfully fabricating the AKROD prototype using the Gear Bearing Drive (GBD) it was concluded that the GBD required further R&D in order to work for the knee rehabilitation application as discussed in Section 4.9. Therefore it was decided that in parallel to the development work on the GBD, the brace would be redesigned using a conventional actuator technologies for the following reasons:

a) Overcome the shortcomings of the previous brace design (as discussed in Section 4.9)

b) Develop the control architecture and algorithm.

Except for minor changes, the framework of the control algorithm developed using the conventional DC motor would be applicable to the sensored GBD. Once the GBD has been redesigned with Hall Effect sensors and is under servo control it will behave in
same manner as a conventional motor. Therefore from a control algorithm view point, the conventional DC motor can be easily be replaced by the sensored GBD.

From the conclusions reached from the previous design, it was decided that the existing brace would be redesigned to meet the following requirements

a) Provide mounting for the new actuator without compromising the design requirements.

b) Include mechanical stops to prevent hyperextension. The stop location should be adjustable between 0° and -5°.

c) Reduce the mechanical play in the yolk and brace interface.

d) Improve the existing disengagement mechanism.

Based on the above requirements and targeted specifications, a brushless DC motor and a planetary gearbox were acquired to replace the GBD for actuating the brace.
5.2 Design Specifications

The DC motor and the actuator used in this design were chosen based on the original design requirement of 50% robotic assistance. Shown below are the detailed specifications of the AKROD using the brushless DC motor and the gearbox.

<table>
<thead>
<tr>
<th>Specifications</th>
<th>Values</th>
</tr>
</thead>
<tbody>
<tr>
<td>Brace</td>
<td></td>
</tr>
<tr>
<td>% of robot assistance</td>
<td>50%</td>
</tr>
<tr>
<td>Range of Motion</td>
<td>-5° to 90°</td>
</tr>
<tr>
<td>Total Weight</td>
<td>9 lbs (4 kg)</td>
</tr>
<tr>
<td><strong>Brushless DC Motor</strong></td>
<td></td>
</tr>
<tr>
<td>Manufacturer</td>
<td>MCG</td>
</tr>
<tr>
<td>Model</td>
<td>IB23000</td>
</tr>
<tr>
<td>Motor Type</td>
<td>Brushless DC</td>
</tr>
<tr>
<td>Continuous Rated Torque</td>
<td>0.286 Nm</td>
</tr>
<tr>
<td>Peak Rated Torque</td>
<td>0.572 Nm</td>
</tr>
<tr>
<td>Rated Speed</td>
<td>6000 RPM</td>
</tr>
<tr>
<td>Torque Constant</td>
<td>0.079 Nm/A</td>
</tr>
<tr>
<td>Current at Cont. Stall Torque</td>
<td>3.01 A</td>
</tr>
<tr>
<td>Current at Peak Torque</td>
<td>12.75 A</td>
</tr>
<tr>
<td>Weight</td>
<td>0.662 kg</td>
</tr>
<tr>
<td>Inertia</td>
<td>0.00001342 kg-m²</td>
</tr>
<tr>
<td>Length</td>
<td>88.8 mm</td>
</tr>
<tr>
<td>Width</td>
<td>57.2mm</td>
</tr>
<tr>
<td>Height</td>
<td>57.2 mm</td>
</tr>
</tbody>
</table>
### Gearbox

<table>
<thead>
<tr>
<th><strong>Gearbox</strong></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Manufacturer</strong></td>
<td>Anaheim Automation</td>
</tr>
<tr>
<td><strong>Model</strong></td>
<td>GBPH-0602-NP-050</td>
</tr>
<tr>
<td><strong>Gear Type</strong></td>
<td>Planetary</td>
</tr>
<tr>
<td><strong>Gear Ratio</strong></td>
<td>40:1</td>
</tr>
<tr>
<td><strong>Rated Output Torque</strong></td>
<td>47 Nm</td>
</tr>
<tr>
<td><strong>Weight</strong></td>
<td>1.40 kg</td>
</tr>
<tr>
<td><strong>Inertia</strong></td>
<td>0.48 kg –cm$^2$</td>
</tr>
<tr>
<td><strong>Length$^3$</strong></td>
<td>62 mm</td>
</tr>
<tr>
<td><strong>Width</strong></td>
<td>60 mm</td>
</tr>
<tr>
<td><strong>Height</strong></td>
<td>60 mm</td>
</tr>
</tbody>
</table>

---

### 5.3 Assembly

Similar to the previous section, the overall assembly of design is briefly discussed first and then followed by detailed explanation of the individual components. The final design of the Brushless DC Motor based AKROD is as shown in Figure 51.

As seen in Figure 51 the Gear Bearing Drive (GBD) has been replaced by the DC motor and the gearbox, thus maintaining the original design requirement of the actuator placement. The DC motor and the gearbox is located in front of the patient’s thigh above their patella thereby placing it closer to their centre of mass to minimize the inertia effect of the actuator. The output shaft of the gearbox has a mechanical link that interfaces with the push bar assembly to transmit the torque from the actuator to the yoke and the lower brace to the move the patient’s knee joint via the load cell. Also included in this design

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$^3$ Excluding the output shaft
are the mechanical stops that will prevent hyperextension in the patient’s gait by preventing motion to the gear box link (discussed in detail in the sections to follow).

Figure 51: Final assembly of the Brushless DC Motor based Active Knee Rehabilitation Orthotic Device (AKROD)

The disengagement mechanism is attained by using easily removing the fastener that connects the clevis and the push bar. Bearing have incorporated where necessary to reduce play in the system.
Similar to the previous version, all the new components (unless otherwise mentioned) of this design were machined from aircraft grade 7075 Aluminum. Its excellent physical properties of higher tensile strength (>500 MPa) and lightweight (density 2.81 g/cm) made it an ideal choice for material selection.

Overall this brace design has fulfilled most the design requirements that were considered necessary for the testing and evaluating the device on the patient’s leg in a clinical environment.

### 5.4 DC Motor and Gearbox

The DC motor and the gearbox for this version of the AKROD was selected such that it can output the torque level required as per the original design requirements yet at the same time add least amount weight and inertia to the overall system.

![Model IB23000 Brushless DC Motor](image)

**Figure 52: Model IB23000 Brushless DC Motor**

The DC motor chosen is a NEMA23 type brushless motor which can output 0.275 Nm of continuous torque and 0.572 Nm of peak torque. These specifications suited for the AKROD application because of the torque requirements generated/absorbed at the knee
During the gait cycle of a healthy adult human, there is mostly continuous torque with certain peak torque occurring for a very short amount of time. Hence the motor was suitable when used in combination with gearbox that has a 40:1 reduction ratio. Another advantage of the motor includes a built-in digital incremental encoder. This is very useful for control algorithm implementation as it gives a direct feedback of the motor position with the least amount of electrical noise during data acquisition.

The gearbox is a planetary type gearbox that has a very low backlash (6 arcmin) and is also back drivable which was an important decision criteria for the implementation of the control algorithm. Also the maximum allowable torque exceeded the torque at the knee joint.

Figure 53: 40:1 Ratio Planetary type Gearbox
5.5 Brace Components

5.5.1 Actuator Mount

The gearbox is mounted on a part called as actuator mount. One end of the support is fastened to the medial side of the brace and the other end to the actuator support and the lateral brace. It provides the rigidity required to maintain the alignment between the lateral and the medial brace. The actuator support is made of SLA-40 plastic material using the 3-D rapid prototyping machine.

It is also strategically made hollow (except where the fasteners from the brace mates at either ends) to reduce the material and weight of the part. Analysis had showed that it will be able sustain the torque loading generated by the gearbox.

Figure 54: Actuator Mount to support the DC Motor and Gearbox
5.5.2 **Actuator Support**

In addition to the actuator mount, the DC motor and the gearbox is provided with an addition components to support the weight of the system and reduce the stress on the actuator mount. The combination of the actuator support and the actuator mount in the design allowed the use of the SLA-40 plastic material for the actuator mount; thereby reducing the weight of the individual components.

![Actuator Support diagram](image)

*Figure 55: Actuator Support to reinforce the DC motor and gearbox mounting to the brace*

The actuator support mated with the lateral side of the brace using socket head screws as shown in Figure 55.

5.5.3 **Gearbox Link**

The torque from the output shaft of the gearbox to the push rod assembly is transferred using gearbox link. One end of the gearbox link is press fitted into the output shaft of the
gearbox and the other to the clevis of the push bar assembly as shown in Figure 56 and Figure 57. Through hole cut-outs have been strategically incorporated in order to remove material for weight reduction.

Figure 56: Gearbox Link to transfer torque from the output shaft of the gearbox

The gearbox link and the clevis interface includes ball bearing as seen in Figure 57. Flanged ball bearings were press fitted into the gearbox link and the assembly is fastened using shoulder screw. This will eliminate play in the system, thereby allowing minimum loss in the torque transfer from the actuator the lower brace assembly.

Figure 57: Gearbox link interface with the push bar assembly
5.5.4 Mechanical Stop

One of the main design criteria to be fulfilled for this version of AKROD was to incorporate a method of mechanically preventing the patient from hyperextension of their leg. In other words, the brace had to have a “hard stop” that will prevent it from further movement. Additionally it was also required that the position of the “hard stop” be adjustable between 0° and -5° to suit the needs of the varied range of patients. The hard stop would have to be easy to access to make adjustments and also be comfortable to use both for the patient and the physical therapist.

After brainstorm various options, it was decided that the mechanical stop should directly stop the rotation of the output shaft rather than motion of the brace. The reason behind it is that if the brace is prevented from rotating using a hard stop, the torque from the actuator will continue to force the brace to rotate and this will cause large amount of stress, which is undesirable.

Shown in Figure 58 is the mechanical stop that is fastened to the gear box. The L-shaped mechanical stop has two nylon tip screws positioned such that the tip of the screw lies along the circular path of the rotation of the gear box link thereby preventing undesired motion of the patient’s knee joint.
As shown in Figure 59, the hex head screws can be easily adjusted by the patient or the therapist such that tip will contact with the gearbox link to obtain the desired hyperextension stop (ranging between 0° and -5°). Nylon tip have been specifically used to reduce/eliminate the noise if and when the gearbox link comes in contact with it.
5.5.5 Push Bar Assembly

The push bar assembly is fundamentally similar to the GBD based AKROD design. However, certain modifications had been made to incorporate the following:

a) Improve the disengagement mechanism.

b) Bearing interface with the actuator/gearbox link.

As seen in Figure 60, the push bar assembly consists of three main components – clevis, push bar, and inner connecting rod. Each of them, including their functionality, is described below.
**Clevis**

The clevis connects the push rod and the gearbox link. As described in Section 5.5.3, the interface between the clevis and the gearbox link is made by means of press fit ball and roller bearings. The push bar (along with the inner connecting rod) and the clevis are fastened using a screw. This screw essentially allows for disengagement between the gearbox link and the lower brace assembly. When completely disengaged, the inner connecting rod is free to move along the length of the push bar thereby not transmitting torque from the actuator to the yoke.

**Push Bar**

When the screw on the clevis is engaged the push bar and the inner connecting rod move together linearly to transfer torque to the yoke and lower half of the brace assembly.

**Inner Connecting Rod**

The inner connecting rod extends into the push bar to engage with the screw at the clevis. On the lower end, it has threaded hole to allow for the load cell to be mated.

**5.5.6 Yolk**

From the GBD based AKROD design it was concluded that the play in the yolk interface with the lower brace has to be reduced. Therefore, in this version of the brace the yolk design was modified to include ball and roller bearings.
As seen in Figure 61, bearing have been press fit into the mating location of the lateral and the medial brace and is fastened using the shoulder screws. Using of bearings (instead of pins mechanism as used in the previous design) allowed for a smooth and frictionless motion of the yoke with respect to the brace thereby eliminating any mechanical play in the system. Overall torque transfer takes place with minimal losses.

## 5.6 Other Components

To reiterate, all the components (mechanical and electrical) not discussed in the preceding section has remained the same from the previous version without undergoing any modifications or changes. The new parts have either been modified or redesigned to accommodate the actuator (DC motor and gearbox) or to overcome the shortcoming that were previously identified.
5.7 Final Prototype

The final prototype of the Brushless DC Motor based AKROD was built as described in preceding sections. All the new metal parts were fabricated at WGI, MA and the SLA plastic parts were made using the rapid prototyping machine in the Biomedical Mechatronics Laboratory, Northeastern University. Shown in Figure 62, Figure 63 and Figure 64 is the final assembled device as well as being worn on a human leg.
Figure 62: Final Prototype of the Brushless DC Motor Based AKROD on a human leg

Figure 63: Front and Side View of the Brushless DC Motor Based AKROD
Figure 64: Hyperextension being prevented by the mechanical stops at the heel strike of the gait cycle
CHAPTER 6 CONTROL SYSTEMS

6.1 Overview

In robot based rehabilitation orthotic device that involves the human gait cycle, a state machine has to be implemented in order to control the behavior of the device. Human gait cycle is a repetitive process that is divided into stance and swings states. With the goal of gait rehabilitation, in order to achieve normal walking capabilities the orthotic device assisting the patient has to be controlled such that a different modality is implemented at any given time depending on the stage of the gait cycle; hence a need for state machine controller.

From a rehabilitation point of view, the overall goal of AKROD is to design a device to restore stroke patient’s to ambulate with an efficient and clinically desirable knee biomechanics. In the medical field, it is agreed that patients with mild-to-moderate impairment and good residual functional capability, restoring a normal gait pattern is best course of action. With this in mind, AKROD has to have capability of

- Aiming to restore the knee flexion/extension angle pattern
- Aiming to restore the knee flexion/extension moment pattern

In addition to controlling the position and moment of the patient’s knee joint, another technique widely used in rehabilitation known as the impedance control also has to be implemented. Detailed discussion provided in Section 6.5.
In order to move towards the development of state machine for the AKROD application, the control system must have the following:

- Low level controllers that mimic normal pattern of motion during ambulation at the knee joint
  - Position Controller
  - Torque Controller
  - Impedance Controller
- Local sensing to determine the exact stage of the gait cycle for appropriate low level controller type implementation

The overall architecture of the control system is as shown in Figure 65.

![Figure 65: Control System Architecture of AKROD using State Machine](Image)

As seen, the control system of the AKROD consist of the three low level controllers (position, torque and impedance) that will be implemented depending of the stage of the gait cycle that the patient is currently in. Data from the sensors will determine the exact
position of the knee joint, torque at the knee joint as well as the phase of the gait cycle. This will be fed back to the state machine that will determine the type of the controller.

Chapter CHAPTER 7 discusses the low level controller that was developed in detail. The algorithm was developed using the National Instrument’s LabVIEW Real-Time Graphical User Interface program. Additional information about LabVIEW Real-Time System provided in Appendix B
6.2 Reference Trajectories

To test and develop the control algorithm of the AKROD, it was required to obtain trajectory data for the human knee joint. This information was acquired by testing healthy human subjects at Motion Analysis Laboratory in Spaulding Rehabilitation Hospital, Boston.

Figure 66, Figure 67 and Figure 68 shows the plots of the knee joint trajectories of 70 adult healthy subjects that were tested in the Motion Analysis Laboratory to establish normative data for clinical gait evaluations. This data was used as references for establishing design criteria as well as for controller development and evaluation.

Figure 66: Normative Knee Angle Data
Figure 67: Normative Knee Velocity Data

Figure 68: Normative Knee Torque Data
6.3 Position Control

The initial goal from control algorithm viewpoint towards the development of impedance control and state machine was to implement conventional position controller. Figure 69 shows the block diagram of the trajectory tracking controller.

A standard PD controller is used to control the equilibrium position $\theta_d(s)$ of the knee joint angle. Using the reference trajectory data obtained from the healthy subject, the desired position (function of time) is input into the algorithm. The input voltage $V_{\text{amp}}(s)$ commanded to the amplifier which is given by the equation:

$$V_{\text{amp}} = K_p (\theta_d - \theta) + K_D s (\theta_d - \theta),$$

Where $K_p$ is proportional term and $K_D$ is the derivative term of the controller.

In this described controlled system, the amplifier is set to current/torque controlled mode to produce the desired force on the motor. The current controlled mode is chosen so that the amplifier will send current command to the motor irrespective of the load at its end. The amplifier takes in the voltage $V_{\text{amp}}$ and outputs a current $I_m$ to be delivered to the
motor to drive the load to the desired position \( \theta(s) \). Current \( I_m \) is proportional to the torque at the motor given by the equation

\[
\tau_m(s) = K_T I_m
\]

where \( K_T \) is the torque constant specific to the motor used in the system.

The position feedback in the above control scheme is provided by the incremental encoder on the Brushless DC motor. The data from the incremental encoder is used to translate into the instantaneous angle at the intersection of the lower and the medial brace. This information can be verified by the angle read from the potentiometer that is already included in the design.

The desired knee trajectory is time scalable to adjust it to meet the requirements of the patient’s individual gait requirements.

This above described position controller algorithm was evaluated on a test bed to quantify the performance of the Brushless DC motor that was chosen for the AKROD design. The test bed and the results are discussed in detail in Chapter CHAPTER 8.
6.4 Torque Controller

Torque controller is the one of low level controllers. The Proportional (P) controller is used to send voltage command to the amplifier which in turn outputs current commands to the motor. The current command is equivalent to the torque used to drive the load at the motor’s end. The load cell measures the force and using inverse dynamics the torque at the knee joint on the brace is calculated.

A simple 2nd low pass filter was incorporated into the controller design to eliminate noisy signal seen from the load cell. The cut-off frequency for the filter was chosen to be 500Hz which is sufficiently larger than the frequency of the normal human gait cycle.
6.5 Impedance Control

The next progressive step in the controls algorithm development process was the development of impedance control.

Impedance control is traditionally used in applications that involve robot-environment interaction. Fundamental principle of impedance control is to recognize the robot’s task as a relationship between its position and the force [41]. For the AKROD’s application, we are dealing with robot (AKROD) and environment (patient) interaction. Therefore an impedance controller is a logical implementation in the control algorithm. It is also important to note that the patient using this device will/might have voluntary motor control to a certain extent. Therefore the idea is to motivate the patient to influence their movement pattern but also at the same time provide guidance along the gait trajectory by allowing robot compliance [42]. In other words from a controls view point, the impedance controller allows the patient to deviate from the prescribed trajectory and experience stiffness depending on the level of impedance chosen, unlike pure position controller that does not allow any deviation by the desired trajectory.

Shown in Figure 71 is a generic schematic of an impedance control. This type of controller is also known as a force based impedance control and is recognized with other names as well. For this document, a force based impedance controller will simply be referred to as “impedance control”.

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The inner feedback loop is force based and the outer feedback loop is position based. The relationship between the change in force and change in position is referred to as mechanical impedance $Z$ which drives the level of robot compliance i.e. how stiff/soft the end point of the robot manipulator can be.

Using the above discussed schematic, an impedance control scheme was developed for the AKROD application as shown in Figure 72.

The inner force loop is measured using the load cell between the push bar assembly and the yoke. Using the information from the load cell, the instantaneous torque at the knee joint on the brace is calculated. The position error of the measured knee joint angle is used to calculate the torque $\zeta_{in}$ that forces the patient’s knee to follow the desired
trajectory. A PD type controller is used for the impedance. Using this, the torque $\zeta_{in}$ is represented by the following equation:

$$\Delta \tau = K_p(\theta_d - \theta) + K_D s(\theta_d - \theta)$$

Where $K_p$ is proportional term and $K_D$ is the derivative term of the controller.

The force on the load is used to calculate the torque $\zeta_{load}$ using inverse dynamics. Referring to the Figure 72, the torque $\Delta \zeta$ is scaled using a factor $G$ to minimize the torque error $\zeta_{in} - \zeta_{load}$. Summing the feedforward loop $\zeta_{in}$ and the $\zeta_{comp}$, the total torque is then used to move the AKROD.

In the above equation, the mechanical impedance is given by

$$Z = K_p + K_D s$$

Smaller $Z$ allows for greater position error, thus more robotic compliance. For a patient’s this means that he or she is able to deviate from the prescribed trajectory with a greater ease compared to a higher value of $Z$. It is also to be noted that the system becomes unstable for greater values of $K_p$ and $K_D$ terms as the gain of the outer loop increases with increasing values of $K_p$ and $K_D$.

The scaling factor $G$ in the above torque controller is essentially a P-controller. When $G$ is set to 0, the output of the torque controller is 0 and therefore the entire control system now behaves as a “simple position controller” [43]. For $G > 0$, allows for compliance to begin as the fraction of the torque error in addition to the $\zeta_{in}$ is commanded to the plant. It is to be noted that higher values of $G$ (for a chosen impedance controller) will lead to instability in the system due to increased gain from the inner force feedback loop. Further
testing and evaluation is required to identify the values $K_p$, $K_d$ and $G$ for which the system will exhibit stability.
CHAPTER 7 CONTROL ALGORITHM

IMPLEMENTATION

The control design discussed in Chapter CHAPTER 6 was implemented using LabVIEW Graphic User Interface System. In this chapter the algorithm implementation, hardware and software is discussed in detail.

Shown in Figure 7.3, is the schematic of the interaction that takes place between the AKROD (mechanical system) and the control architecture. The controller essentially uses two hardware components, namely the host computer and the target computer. The user (physical therapist or the patient) can input a desired trajectory and control parameter using the host computer. And the host computer communicates to the target system by means of a shared variable engine. The sensor data from the AKROD is sent to the target. The target system then commands an appropriate action to be taken based on the input from the host and the controller settings.
7.1 Hardware

The control system of AKROD is facilitated using two computer system – Host and Target. As shown in Figure 74, host computer is used for user interface and data logging. This is where the user is able to input the desired trajectories, adjust the gains of the controller, and view the various responses (position, torque etc.) of the system in real-
time. The target computer is the dedicated controller of the system. It runs all of the critical tasks like data acquisition, controls, and other processes. Using a dedicated real-time system results in tasks that more stable and the overall system has a robust operation. The communication between the host and the real-time target takes places through Gigabit Ethernet connection.

![Figure 74: Host and Target System for Controller Implementation](image)

The data from the sensors (incremental encoder on the DC motor, load cell etc.) is acquired using National Instrument’s PCIe-6259, M series high-speed multifunction data acquisition board. A BNC terminal box, BNC 2120 from National Instrument is used to transfer the data signal to and from the PCIe 6259 data acquisition as shown in Figure 75.
The amplifier used in the controller design is a Xenus servo amplifier manufactured by Copley Controls Inc. The digital amplifier has the options to be programmed to operate in three different modes – position mode, velocity mode and current/torque mode.

As previously mentioned, a current controlled mode is chosen for the controller design. Shown in Figure 77 is the built-in current controller used to command the current to drive the motor load. It consists of a current limit that receives a “desired” current command $I_m$. (This current command is determined by a multiplication factor of the analog voltage...
command $V_{amp}$ i.e. output of the LabVIEW RealTime System on the target computer). The “desired” current is the equivalent torque that is produced at the motor given by the equation

$$\tau_m(s) = K_T I_M$$

where $K_T$ is the torque constant of the motor.

The limiting stage accepts the current command, applies limits and passes a limited current, subtracts the actual current to produce an error signal. The error signal is then processed using a PI (Proportional Integral) controller gains to produce a command. The command is then forwarded to the amplifier’s power stage to drive the load on the motor.

Combining the various pieces of the control system and mechanical system discussed in the preceding sections, the following section attempts to explain the implementation of the position controller and impedance controller that was proposed earlier.

Figure 77: Current controller of the Xenus servo amplifier
7.2 Position Control

In the position control mode, the user inputs a desired trajectory $\theta_d(s)$ into the LabVIEW GUI using the host computer. The desired position $\theta_d(s)$ is forwarded to the summing junction in the LabVIEW Real Time (RT) system.

Under the LabVIEW RealTime (target) level, the summing junction takes the $\theta_d(s)$ and subtracts the actual position $\theta(s)$ to produce the position error signal. The actual position is determined from the incremental encoder on the brushless DC motor. The error signal is then processed through the Proportional Derivative (PD) controller using the gains chosen by the user at the host computer level to produce an analog voltage command (equivalent to a certain torque $\zeta_m$ to drive the motor load).
The amplifier converts the $V_{amp}$ to an equivalent current $I_m$ ($10V = 9.94A$ was chosen as the conversion factor) and processes it using its own PI controller to produce a current command which is then amplified to drive the knee brace to the desired position.

In the AKROD, the brace move the patient’s knee joint to the desired position which is read by the incremental encoder. This position is the actual position $\theta(s)$ that is used in the position feedback loop to determine the position error. This process continues until the brace has reached the desired position; in other words until the position becomes zero.

The above described position controller was tested to evaluate and quantify the performance of the brushless DC motor on the testbed setup.
7.3 Impedance Control

As discussed earlier, the goal of the impedance control is to control the relationship between the end point contact force of the robotic manipulator and the deviation between the desired and actual position of the manipulator. Shown in Figure 79 is the schematic of the impedance controller that was implemented for the AKROD application. It consists of two feedback loops – the inner force control loop and the outer position control loop. Using these, the impedance control law aim at controlling the relationship between the torque and position is given by

\[ Z = \frac{\Delta \tau}{(\theta_d - \theta)} \]

where \( Z \) is defined as the mechanical impedance. In the above scheme, the impedance controller is a PD controller and is realized by

\[ Z = K_p + K_ds \]

From the above equation, it is seen that depending on the gains of the impedance controller will allow the amount position deviation error. Therefore higher impedance will lead to smaller deviation and vice versa.

For implementation, at the host level the user inputs the desired trajectory into the LabVIEW GUI and adjusts the desired PD gains of the impedance controller and the scaling factor \( G \). The desired position \( \theta_d(s) \) is then forwarded to the RealTime target system.
Figure 79: Impedance Control Implementation
In the LabVIEW RT system, the summing junction takes the $\theta_d(s)$ and subtracts the actual position $\theta(s)$ to produce the position error signal. The position error signal is then processed through the PD controller to produce an analog voltage command $V_{in}$. In the next summing junction, $V_{in}$ is subtracted by the variable $V_{load}$.

$V_{load}$ is obtained from the inner force loop equivalent to the signal obtained from the load cell. The force signal from the load is multiplied by the moment arm (normal distance between the load cell and the centre of the polycentric hinge joint on the brace arm) to obtain the torque delivered by the actuator. Assuming the system is 100% efficient, torque at the knee joint brace arm is torque at the output of the gear box. There it torque is dived by gear ratio factor to obtain the torque at the motor. It is then converted to equivalent current (dividing by torque constant) and then to equivalent voltage command $V_{load}(s)$. The voltage error $V_{in} - V_{load}$ is the equivalent to the torque error $\Delta \zeta$. Voltage is then processed through the P-controller to output a signal $V_{amp}(s)$ which scaled by a scaling factor $G$. The signal $V_{amp}(s)$ is then forwarded to the amplifier. The amplifier converts the $V_{amp}$ to an equivalent current $I_m$ ($10V = 9.94A$ was chosen as the conversion factor) and processes it using its own PI controller to produce a desired current command which is then amplified to move the motor load.

In the impedance control scheme, the patient has the ability to deviate from the desired trajectory and this therefore results in a compliant robotic manipulator. As mentioned earlier, the extent of deviation depends on the gain chosen for the impedance controller as well as the scaling factor $G$. At the far end of the schematic, patient and the therapist is able to monitor the performance by comparing the input trajectory and actual trajectory in the LabVIEW GUI at the host level.
CHAPTER 8 CONTROLLER EVALUATION AND TESTING

8.1 Testbed Setup

For the preliminary testing and evaluation of the control algorithm, a test bed was designed and setup to simulate the functionality of the AKROD. The goal of the test bed was to

- Mimic the knee joint angular movements
- Quantify and evaluate the performance and response characteristics of the Brushless DC Motor and gearbox
- Debug and optimize the control algorithm code before conducting any testing on the AKROD prototype

The testbed setup to conduct the preliminary testing is as shown in Figure 80. It consists of a pulley system that interfaces with the output shaft of the gearbox. Using rope string, weights are attached to system. The load cell is included between the pulley and weight itself. The rotary motion of the pulley is equivalent to the flexion/extension movement of the human knee joint. In the existing setup, a constant torque (load) is observed by the motor because the weight attached to the pulley is always tangential. The maximum torque that can be applied to the system is 10Nm, which is within the working range of the AKROD.
The maximum torque chosen here was limited by two factors

- The pulley was fabricated on the rapid prototyping machine. It was limited to a diameter of 12 inches.
- The weight attached had to be within the reasonable limit for a safe environment to conduct the tests

Taking the above into consideration, the pulley was designed to have a radius of 4.5 inches (114.3 mm) and maximum weight used a 20lb dumbell. Based on this, a maximum of 10 Nm of torque can be simulated on the overall system.

The following assumptions were made for the above mentioned testbed
• The flexion/extension movements is directly caused by the actuator. In other words, the mechanical linkage equivalent to the pushbar assembly and yoke is absent from the testbed

• The link between the pulley and the weight is stiff. The disturbances due flexibility in the wire rope is ignored.

8.2 Test Results

One of the first steps towards the development of the control system, it was essential to characterize and quantify the chosen actuator by performing time response analysis. The following sections describe results of the tests conducted on the testbed to understand the closed loop performance of the position controller described in Section 7.2

8.2.1 Step Input Response

The first set of test conducted on the position controller was step response for a load of 0lb, 5lb and 20lb. A maximum of 20lb was chosen such that the system experiences a maximum torque of about 10 Nm. The step inputs were chosen to be 20°, 40° and 60° of rotation. The maximum step input of 60° is along the lines of the peak flexion made the knee joint during a normal gait cycle. However it is unlikely that that the AKROD will experience such a large step during regular working condition.

The gains of the controller were tuned experimentally such that there was no overshoot in the response of the system at any given weights and the steady state error was maintained at less than 0.7°.
Figure 81, Figure 82 and Figure 83 show the step inputs and response of the system at the specified weight/torque.

Figure 81: Step Input Response for 0 lb weight
Figure 82: Step Input Response for 5lb weight

Figure 83: Step Input Response for 20lb weight
8.2.2 Sine Input Response

Using the gain tuned in the step response, the system was subjected to sine inputs at frequency of 0.9 Hz. This frequency was chosen to match the period of gait cycle of normal human adults which last for approximately 1.2 sec. The amplitude of the sine wave was chosen to be 20°, 40° and 60°.

Figure 84, Figure 85 and Figure 86 shows the sine response of 0lb weight\textsuperscript{4}.

\textsuperscript{4} Sine response has separated in individual plots for better clarity.
Figure 85: Sine Response for 0lb weight; Amplitude: 40°

Figure 86: Sine Response for 0lb weight; Amplitude: 60°
Figure 87, Figure 88 and Figure 89 show the sine wave response for 5lb weight at the specified gains.

Figure 87: Sine Response for 5lb weight; Amplitude: 20°
Figure 88: Sine Response for 5lb weight; Amplitude: 40°

System Parameters
- Weight: 22.2 N (5 lb)
- Torque: 2.5 Nm

Gains
- Kp = 0.7
- Kd = 0.03

Sine Wave
- Amplitude = 40°
- Frequency = 0.90 Hz
- Phase Lag = -1.62°

Figure 89: Sine Response for 5lb weight; Amplitude: 60°

System Parameters
- Weight: 22.2 N (5 lb)
- Torque: 2.5 Nm

Gains
- Kp = 0.7
- Kd = 0.03

Sine Wave
- Amplitude = 60°
- Frequency = 0.9 Hz
- Phase Lag = -0.972°
Figure 90, Figure 91 and Figure 92 shows the sine wave response for 20lb weight (Torque ≈ 10Nm). It is seen from Figure 90 that the system did not track as well as it did at low torque values. Hence for experimental purposes, the existing PD (Proportional Derivative) controlled was substituted with a PID (Proportional Derivative Integral) controller. The Integral gains were experimentally tunes to 0.003 until the tracking demonstrated an improved response. This is evident from the sine wave response with amplitudes of 40° and 60° (Figure 91 and Figure 92).
Figure 91: Sine Response for 20lb weight; Amplitude: 40°

Figure 92: Sine Response for 20lb weight; Amplitude: 60°
8.2.3 Knee Trajectory

In next series of test, the controller was subjected a position input (or desired input) to track the knee angle trajectory of normal human gait cycle. The normative data obtain from the Spaulding Rehabilitation Hospital (Section 6.2) was linearly interpolated from 100 points to 5000 points using a LabVIEW program. In the LabVIEW RealTime System the while loop rate was set at 5000 Hz frequency. Therefore it is essential that it has a large number of input data points (here 5000 data points); resulting in a smooth input knee trajectory curve. The control system was modified such that each gait cycle simulated the normal walking pattern. The test with 0lb and 5lb weight was conducted for a gait cycle lasting 1 sec and for the test with 20lb weight, each gait cycle lasted for 2 sec. Figure 93 and Figure 94 shows the knee angle trajectory response for 0lb and 5lb weight.
Figure 93: Knee Angle Trajectory Tracking for 0lb weight

Knee Angle Trajectory Response

- Angle (degree)
- Time (sec)

System Parameters
- Weight: 0 N
- Torque: 0 Nm

Gains
- Kp = 0.7
- Kd = 0.03

Gait Cycle
- Period = 1 sec

Figure 93: Knee Angle Trajectory Tracking for 0lb weight
Figure 94: Knee Angle Trajectory Tracking for 5lb weight

Figure 95 shows the tracking of knee angle trajectory for 20lb weight. It can be seen that tracking is not similar to tracking with lower weights even with the additional integral controller. The gains of the integral controller ($K_i = 0.003$) was kept unchanged for comparing the trajectory with the sine wave response (Figure 92).
Figure 95: Knee Angle Trajectory Tracking for 20lb weight

Knee Angle Trajectory Response

System Parameters
Weight: 89 N (20 lb)
Torque: 10.15 Nm

Gains
Kp = 0.7
Kd = 0.03
Ki = 0.003

Gait Cycle Period = 1 sec
8.3 Discussion

The method and tests discussed in the above sections using a constant torque was an initial attempt to simulate the motion of the AKROD along the knee joint and also to debug and optimize the LabVIEW code before implementing it on the prototype. The tests provided a method of quantifying the actuator and foresee any problem/issues that might be encountered at a later stage of development.

From the above results we see that the actuator performed very well at lower weights using the PD controller. However at higher weights (here 20 lb) the tracking was not along the same lines as compared to lower weights (here 0lb and 5lb) using the same PD controller gains. Adding an integral component with very small gain (Ki= 0.003) improved the tracking to a large extent. This was expected because the motor is experiencing larger loads and therefore requires additional current to drive it. However additional testing will be required to determine/optimize the gains of the controller for various scenarios that will experienced during normal use of the device.
CHAPTER 9 CONCLUSION AND FUTURE WORK

9.1 Gear Bearing Drive

Upon fabricating the GBD prototype for the AKROD (Chapter CHAPTER 4), it was determined that the outrunner motor had to have a slow speed start up capability to be suitable for this particular application. The existing outrunners currently available in the market are sensorless. Further research led us to believe that a slow start operation can be accomplished if the off-the-shelf outrunner could have Hall Effect sensors installed in them. If successful, then the motor will rely on the sensors to determine its velocity rather than then back e.m.f. It will also provide a stable and robust servo control operation along the same lines as the conventional DC motor.

9.1.1 Current Activities

A Hacker outrunner motor was speced out and installed with Hall Effect sensors as shown Figure 96. Installation of such Hall Effect sensors was a complex task mainly because outrunner motor are not designed for it and hence the most important issue was space constraints and mounting of the sensors. This was partially overcome by using a smaller motor (Medium size) and replacing the rotating can of a larger motor (L size). Mounting of the sensor was done by designing a custom holder using the rapid prototyping machine on which the sensors can be mounted. The mount was then placed on the motor windings without impeding the motion of the rotor.
Once the Hall Effect sensors were successfully mounted, the next task was to be able to drive the motor under servo control. For this purpose, an Accelnet servo amplifier (manufactured by Copley Controls Inc.) compatible with the outrunner specifications was acquired. Using this, we were able to successfully able to control the outrunner motor under “velocity control” mode and “current control” mode. In other words, we had obtained sensored DC outrunner motor. This entire activity now allowed the outrunner motor to have a start-up speed in the range of 500-600 rpm (compared to > 6000 rpm without Hall Effect sensors).


9.1.2 Future Work

The sensored outrunner motor needs to be characterized for various specifications like torque constant, current draw, torque-speed curve and other power characteristics. For this purpose we have designed and assembled a custom dynamometer as shown in Figure 97.

![Sensored Outrunner Motor](image)

Using the dynamometer, we will run a series of test to quantify and evaluate various outrunner motors to determine the most suitable for gear bearing and AKROD application in entirety. Once the correct motor has been chosen, the gear ratio required to design the gear bearing will be determined.
Next step in the process will be to design the gears of the Gear Bearing Drive out of metal. This process will include a series of tasks consisting of design of the gear tooth, material choice and the fabrication of the gears to name a few.

### 9.2 Control System – Future Work

The position controller was implemented and tested to characterize the actuator performance. The next set of tasks will include testing of the impedance controller on the test bed. This will be followed by migrating the actuator to the final prototype to test and evaluate the controller.

In parallel to this, we will also research into methods of determining the torque input by the patient. The existing load cell in the design provides the overall torque at the knee joint due to the actuator. However, another important component of torque is to determine the patient intention – either to assist to motion of the brace or oppose the motion of the brace or do nothing. Patient intention is partially dependent on the stage of the gait cycle and transition from one stage to another within the gait cycle. We will research and establish the optimal method that can help ascertain the above-mentioned data. Next set of tasks will include the development of state machine using the sensing information to ascertain which low-level controller to be used for a given stage of the gait cycle, patient capability and therapist prescription.
APPENDIX A: Virtual Reality (VR) for AKROD

The proposed AKROD's Virtual Reality System will be composed of three hardware components: a head mounted display, a treadmill, and the AKROD as shown in Figure 98. It will also contain two software components: the VR for Calibration and Evaluation (VRCE) and the VR for Motivation and Enjoyment (VRME). The VR scenes will be built using the Panda3D graphics engine.

![Figure 98: Schematics of the AKROD's Virtual Reality System](image)

The VR will be displayed for the patient using a head mounted display (HMD). The HMD will allow the patient to be immersed in the VR environment and receive feedback on their progress. The HMD model chosen for this project is the i-Glasses i3pc (i3PC). The i3PC weighs 8 ounces so it should not fatigue or irritate the patient during
rehabilitation exercises. It connects to a personal computer (PC) via a VGA connector and is capable of stereographic 3D.

For the optimum patient experience while using the AKROD the device will need to be calibrated for each patient. This will include collecting baseline data for the patient’s current gait and assessing how much assistance the device should give the patient. The VR for Calibration and Evaluation (VRCE) interface will use the baseline data to provide feedback to the patient. The VRCE will be using known forces and angles to show a comparison of how the patient is performing compared to a simulation of the correct gait (see Figure 99-LEFT).

Figure 99: Comparison of Patient and Test for the Evaluation and Configuration VR (LEFT); Layout of the VR for Motivation and Enjoyment (VRME) interface (MIDDLE); Possible Virtual Scene for the VRME (RIGHT).
APPENDIX B: DC Motor Outrunner Controller
APPENDIX C: LabVIEW RealTime System

The software of the AKROD runs on a real-time platform, allowing accurate timing characteristics to the system. Regular DAQ hardware running on a general-purpose OS such as Windows cannot guarantee real-time performance since some factors (i.e. programs running in the background, interrupts, and graphical processes) may compromise the performance. In contrast, real-time hardware running on real-time operating systems (RTOS) allows the programmer to prioritize tasks so that the most critical task (such as differentiation or controls) can always take control of the processor when needed. This property enables reliable applications with predictable timing characteristics.

A real-time system consists of a host computer (usually a laptop or standard desktop PC), and a target system that runs the RTOS. The code is developed on the host, and then deployed to the RT target. Communication between the host and RT target is via high speed ethernet. The RT target streams critical parameters back to the host for monitoring. The host displays and saves the streamed data, and also displays the patient and practitioner GUIs. To facilitate cost reductions and portability, future versions of the system can utilise board level RT targets that feature FPGA chips and single board computers. These solutions offer a significant packaging advantage at the expense of flexibility.
Figure 100: RealTime Interface Options

Figure 101: Host - Target Communication
Figure 102: Loop Iteration Timing variability on Windows OS

Figure 103: Iteration Timing on Windows Operating System
Figure 104: Iteration Timing on LabVIEW Real-Time Operating System
REFERENCES


