DESIGN OF A NOVEL TASK-BASED KNEE REHABILITATION EXOSKELETON
DEVICE WITH ASSIST-AS-NEEDED CONTROL STRATEGY

A Thesis by
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DESIGN OF A NOVEL TASK-BASED KNEE REHABILITATION EXOSKELETON DEVICE WITH ASSIST-AS-NEEDED CONTROL STRATEGY

The following faculty members have examined the final copy of this thesis for form and content, and recommend that it be accepted in partial fulfillment of the requirement for the degree of Master of Science, with a major in Mechanical Engineering.

________________________________________
Yimesker Yihun, Committee Chair

______________________________________
Hamid M. Lankarani, Committee Member

________________________________________
Jaydip Desai, Committee Member
DEDICATION

To my advisor, Dr. Yimesker Yihun
Who have been consistent source of inspiration and knowledge for me
and
To all My Family and Friends
ACKNOWLEDGEMENTS

The completion of this thesis is a great endeavor for which I am obliged to many people whose contribution are sincerely appreciated and gratefully acknowledged. I am highly indebted to my advisor, Dr. Yimesker Yihun, for his guidance, motivation and support throughout the course, without him it might have been impossible task for me. His commitment towards the work motivated me to follow his path and complete the thesis work on time.

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At last but not the least, I am thankful to all my family and friends for their endless support.
ABSTRACT

This thesis aims to design a novel task based knee rehabilitation exoskeleton device through kinematic synthesis. In contrast to prevailing research efforts, which attempt to mimic the human limb by assigning each human joint with an equivalent exoskeleton joint (e.g. a hinge joint for the elbow and knee), this thesis provides an alternative systematic approach for the design of exoskeletons to assist the complex 3D motions of the human Knee. With this method, it is not necessary to know the anatomy of the targeted limb, but rather to define the motion of the exoskeleton segments based on its point of attachment to the limb. Good alignment is often difficult and the distances between joints must be adjusted to accommodate the variety of human size. Furthermore, attempting to align each robotic joint axis with its human counterpart assumes that the position of the axis can be accurately known, and that such a fixed axis exists for the range of motion of the joint or set of joints, which is not always the case. In human-exoskeletons synergy, especially in industrial settings and rehabilitation applications, due to the repetitive and strenuous nature of the task, the fit, comfort and usability of these exoskeletons are important for the safety of the user and for the automation of the task. Improper fitting may lead an exoskeleton to move in a way that exceeds the range of movement of the human body and tear muscle ligaments or dislocate joints.

In this thesis, to study the motion of the desired clinical trajectories of the human knee, the state-of-the-art of motion capture and data analysis techniques are utilized. The collected experimental kinematic data is used as an input to the kinematic synthesis. Parallel mechanisms with single degree-of-freedom (DOF) are considered to generate the complex 3D motions of the lower leg. An exact workspace synthesis approach is utilized, in which, the parameterized forward kinematics equations of each serial chain are to be converted to implicit equations via elimination. The implicit description of the workspace is made to be a function of the structural parameters of the serial chain, making it easy to relate those parameters to the motion capture data. A prototype of the mechanism has been built using 3D printing technology. And an Electromyography (EMG) signals and Force sensing resistors (FSR) are utilized to implement an assist as needed controller. The EMG signal is captured from the user leg and force sensing resistors (FSR) are applied at the attachment point of the exoskeleton and the leg, this helps to get the amount of force applied by the exoskeleton to the leg as well as for recovery tracking. The assist as needed controller eliminates the need of constant supervision, and hence saves time and reduces cost of the rehabilitation process.
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<tr>
<td>CFRP</td>
<td>Carbon-fiber-reinforced polymer</td>
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<td>DOF</td>
<td>Degree of Freedom</td>
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CHAPTER 1

1. INTRODUCTION

1.1. Background

The knee joint is a major anatomical component of the human body, as it bears the body weight during most of the daily activities, and it is prone to major injuries and illness. In the USA, more than 700,000 people suffer from stroke, 10,000 suffer from traumatic spinal cord injury and 250,000 becomes disabled because of multiple sclerosis each year [1]. These injuries and diseases (like stroke, trauma, inflammation and heart disease) are ever increasing, creating more people with lower leg disability and leave patients with greatly impaired or completely nonexistent muscle function over time, making life for them increasingly more difficult. Some common problems associated with people with disability are

- Inability to sit, perform and walk without assistance
- Losing independence and dependent on walkers, crutches and wheelchairs
- Commercially assistive and rehabilitation devices available, but ineffective and costly [2, 3]

Currently the medical community has yet to find a treatment or cure for many neuromuscular diseases. An option to relieve the symptoms while that medical research continues is rooted in assistive technologies that compensate for the “failure” in the original system. To this end, there has been considerable research directed toward the development of assistive devices to restore lost freedom of motion or limb functionality. For example, there are currently exoskeleton devices being used, largely in physical therapy, to help rehabilitate paralyzed or impaired movement of the human body for victims of strokes and other similar medical problems [4].

1
Studies have shown that the rehabilitation process and therapy techniques are successful in improving the functionality of the rehabilitated part [4-9]. However, there is a different opinion on rehabilitation programs and concepts. Some physicists recommend early mobilization for better recovery while others recommended passive motions only in early stages [10]. Some studies show that specific task based exercises for targeted muscle and ligaments is preferred for recovery speed. Selection of rehabilitation techniques and methods are also play a great role, as incorrect method of rehabilitation can lead to abnormal wear, patellofemoral problems and premature loosening of the muscles [11]. The forces in ligament during the typical rehabilitation exercises (isokinetic and isometric or squat exercises) have been studied in [12], the result shows the type of exercise and force involved have an effect on the recovery speed. Also, studies have shown that the more vigorous is the controlled exercise, the faster the recovery occurs and besides regular exercise, knee rotation involves most of the joint muscles.

The Robotic Exoskeleton is a wearable robotic suit designed to enhance or restore user’s movement abilities by providing support, guidance and protection to the body [13]. And it is believed to be the future of rehabilitation, making it automated, reducing health costs, physician efforts and time [4]. In 1980s, the utilization of exoskeleton for rehabilitation was noticed as it improves walking efficiency [14]. However, the technology was constrained due to several reasons, including unavailability of advanced sensing devices, actuators and high signal processing and feedback techniques [15]. Current developments can provide users with rehabilitation and assistive walking and other motions, but have not quite completely mastered every aspect of an exoskeleton. Improvements can be made in nearly all areas including, but not limited to, the power supply, lifting capability, weight of the frame, fitting and alignment, and how a user controls the exoskeleton etc.
1.2. Literature Review

Exoskeleton can be broadly utilized for assistive, power augmentation, and rehabilitation purpose [16]. The assistive exoskeletons are focused on providing the assistance to people with disability and the elderly to enhance work capability. The load carrying exoskeletons are designed to enhance user’s strength and endurance, which are being extensively used in the military applications for effective gait operation. Rehabilitative exoskeletons are meant for facilitating the recovery of the patient to gain the lost freedom by reducing the pain, swelling, cost and effort through controlled exercise.[17]. It is becoming one of the popular field because of the intrinsic human robot interaction.[4] People suffering from diseases, injuries like spinal cord injury, cerebral vascular accident, or ageing-associated diseases needs rehabilitation devices for performing regular activities as it affects its functionality [18-20]. Muscle expansion and body fluids makes the damaged joints swell to protect it [17, 21]. Human Knee requires long period for the recovery leading to the need of rehabilitation. There are two different interaction scenarios, in first cognitive interaction case; robot is controlled by human with the help of the feedback signal sent by the robot to human. EMG based control prosthesis in which myoelectric signal generate the control algorithm is the example of such kind robots[22]. While in the second case, actuators are controlled through the biomechanical interaction like in exoskeleton based robots which provides the functional compensation of human actions e.g. Gait rehabilitation [23]. Depending on the Rehabilitation principles, rehabilitation robots can be divided into 5 major categories given as [4]:

1. Treadmill gait trainers: In this process, during walking legs and hips are assisted by therapist and overhead harness support some part of body weight. Lokomat, LokoHelp, ReoAmbulator, Ambulation-assisting Robotic Tool for Human Rehabilitation (ARTHuR),
Active Leg Exoskeleton (ALEX), LOPESS (Lower-extremity Powered Exoskeleton), POGO and Pelvic Assist Manipulator (PAM) are some examples of such trainers among which ARTHuR, PAM, POGO are capable of automated gait training.[24]

2. Foot-Plate-Based Gait Trainers are based on the foot plates which is programmed for the specific types of movement or gait pattern. The Gangtrainer GT I, The Haptic Walker, The GaitMaster5 (GM5), The Lower-Limb Rehabilitation Robot (LLRR) are some examples of this kind of exoskeleton.

3. Overground gait trainers allows the patients to move on the ground and are actuated with the servos to replicate the human motion. Some most popular exoskeleton of this kind includes KineAssist, WalkTrainer, ReWalk, HAL, WHERE I-II etc. and these kinds of devices are successfully commercialized.

4. Stationary Gait Trainers are focused for obtaining optimal effect during muscle strengthening, endurance development and joint movement coordination. MotionMaker, Lambda and AIST Tsukuba are some examples of this type of rehabilitation devices.

5. Ankle and Knee Rehabilitation Systems contains stationary type and Active Foot Orthoses. Stationary system is mechanism designed to rehabilitate the ankle and knee without actual walking, some example includes Leg-Robot, NUVABAT, AKROD, Rutgers Ankle, IIT-HPARR etc. Whereas Active Foot Orthoses are actuated exoskeletons. The examples include Powered Gait Orthosis (PGO), Pneumatic Active Gait Orthosis (PAGO), MIT Active Ankle-Foot Orthosis (MIT- AAFO), Ankle Foot Orthosis at the University of Delaware (AFOUD) etc.
Among the different existing rehabilitation exoskeleton technologies, some were successful to take market and attract the intended customers. An example of a successful rehabilitation exoskeleton is the ReWalk. The ReWalk uses electric motors to provide movement at the hips, knees, and ankles. This exoskeleton stores the computer and power supply inside a backpack which is worn by the user. The user controls the exoskeleton via a remote worn on the wrist and can specify to stand up, sit down, or walk. The system weighs 46 pounds, but it carries its own weight so the user only feels the weight of the backpack. In 2014, the FDA cleared the ReWalk Personal for use at home and in the community. Crutches can be used with the ReWalk to provide additional stability [4]. However, there is also a product produced by Rex Bionics that offers similar rehabilitation as well as an extra benefit of not requiring crutches. The system is larger than the ReWalk and the legs of the exoskeleton are linked by a strong hip girdle [25].

C-Brace is the custom carbon fiber designed assistive/rehabilitation purpose leg intended for people with partial paralysis, Spinal injury, post-polio syndrome and post-stroke. it is real time controlled using high processing micro controller, actuated with hydraulic actuator and is equipped with knee angle, knee angle velocity and ankle moment sensor [26, 27]. FreeWalk leg brace is another product from Ottobock designed to improve the walking ability by locking the knee joint when knee-ankle-foot-orthosis reacts to movement [26]. Successfulness of active foot orthosis devices and gait trainers over stationary system shows the mobility need for the rehabilitation and hints for the portable system solution. Briefings on some existing lower leg exoskeleton for the rehabilitation purpose are shown in Table 1.
Table 1
Existing Exo-Skeleton Technologies for Lower Leg Rehabilitation

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<th>Controllers, power system, Actuators and sensors used</th>
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<tr>
<td>EKSO suit</td>
<td><img src="image" alt="Figure 1 ESKO suit" /></td>
<td>-Microcomputers (Wrist Band) -battery powered - 4 motors, one at each hip and joints -Motion sensors</td>
<td>-For paralyzed patients( (rehabilitation) -Cost around 100,000 -Weighs around 23 kg -Joint flexibility/actuator/ Size/power supply problems -Can stand, walk and climb stairs. -Balancing using crutches</td>
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<td>Life suit</td>
<td><img src="image" alt="Figure 2 Life Suit by Monty K Reed" /></td>
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<td>-For rehabilitation purpose -Bulky -Cost around 18,000 -Passive moment of legs</td>
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<th>Model</th>
<th>Image</th>
<th>Controllers, power system, Actuators and sensors used</th>
<th>Remarks</th>
</tr>
</thead>
<tbody>
<tr>
<td>REX[28]</td>
<td><img src="image" alt="Figure 3 REX Bionics" /></td>
<td>- 27 onboard microprocessors manage the actuator systems</td>
<td>- For rehabilitation</td>
</tr>
<tr>
<td></td>
<td></td>
<td>- Lithium-polymer battery (29.6V, 16.5Ah) (sufficient for 1 hour of operations)</td>
<td>- Self balancing</td>
</tr>
<tr>
<td></td>
<td></td>
<td>- 10 customized actuators</td>
<td>- Weighs 84 pounds</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>- Independence in mind</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>- Not adjustable as it is manufactured to fit with an individual measurement</td>
</tr>
<tr>
<td>HAL 5</td>
<td><img src="image" alt="Figure 4 HAL 5 suit" /></td>
<td>- Hybrid control system (CAC + Bio cybernic Control) (Use of small Pc)</td>
<td>- Can be used for Assistive / Rehabilitation</td>
</tr>
<tr>
<td>Hybrid Assistive Limb</td>
<td></td>
<td>- Li ion rechargeable battery</td>
<td>- Lease 2300/month</td>
</tr>
<tr>
<td>[29, 30]</td>
<td></td>
<td>- Bioelectric signal sensor</td>
<td>- Body weight around 20 kg</td>
</tr>
<tr>
<td></td>
<td></td>
<td>- Power sensor</td>
<td>- Carbon and Mg</td>
</tr>
<tr>
<td></td>
<td></td>
<td>- FRF sensor</td>
<td>alloy are used for suit</td>
</tr>
<tr>
<td></td>
<td></td>
<td>- Angle Sensor</td>
<td>- Not good in gravity balancing function</td>
</tr>
</tbody>
</table>
Table 1 (continued)

<table>
<thead>
<tr>
<th>Model</th>
<th>Image</th>
<th><strong>Controllers, power system, Actuators and sensors used</strong></th>
<th><strong>Remarks</strong></th>
</tr>
</thead>
<tbody>
<tr>
<td>ReWalk[4]</td>
<td>Figure 5 ReWalk Rehabilitation Exoskeleton</td>
<td>-Computer and power supply inside a backpack worn by the user</td>
<td>-Rehabilitation (can be used to stand up, sit down, or walk)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>-Electric motors at hip knee and ankles</td>
<td>- System weighs 46 pounds</td>
</tr>
<tr>
<td>The C-Brace® Orthotronic Mobility System</td>
<td>Figure 6 The C-Brace® Orthotronic Mobility System</td>
<td>-Computer/ microprocessor controlled</td>
<td>- For Assistive Rehabilitation (Partial paralysis, spinal injury post-stroke and post-polio syndrome)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>-Hydraulic knee joint controlled with servomotors</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>-No bionic interactions</td>
</tr>
</tbody>
</table>
Table 1 (continued)

<table>
<thead>
<tr>
<th>Model</th>
<th>Image</th>
<th>Controllers, power system, Actuators and sensors used</th>
<th>Remarks</th>
</tr>
</thead>
<tbody>
<tr>
<td>KineAssist[4, 31]</td>
<td>Figure 7 KineAssist Rehabilitation device</td>
<td>-Consists custom designed pelvis and torso harness attached at base</td>
<td>-Electric motors -For Gait and balance training (Rehabilitation) -Real time sensor controlled exercise</td>
</tr>
</tbody>
</table>

These existing challenges show that, many of these are too heavy or bulky for everyday use outside of a physical therapy center [16]. This is partly due to the attempt to align each robotic joint axis with its human counterpart (e.g. a rotational joint for the elbow and knee)[32] (Figure 9 (a) & (b)). As the axis must be external to the human joint, this also adds complexity and weight to the design. Good alignment is often difficult and the distances between joints must be adjustable to accommodate the variances of human sizes. Furthermore, trying to align each robotic joint axis with its human counterpart assumes that the location of the axis can be accurately known, and that such a fixed axis exists for the range of motion of the joint or set of joints, which is not always the case. A clear example of complex kinematic modeling is the Knee, for which precise detection methods such as MRI segmentation show that there is not a fixed rotational axis [33]. The human knee changes its axis during movement (Figure 8 (c)) thus, replicating the joint with a single hinge joint will lead to misalignment and stress on the user.
Figure 8 (a) Exoskeletons designed to align with the human knee joint axes[34], (b) Simplified kinematic models of human arm [35], (c) Changing in helical axis of knee during motion[36]

1.3. Human Knee Kinematics

<table>
<thead>
<tr>
<th>Knee Components</th>
</tr>
</thead>
<tbody>
<tr>
<td>• Femur- Thigh bone</td>
</tr>
<tr>
<td>• Patella- Knee cap</td>
</tr>
<tr>
<td>• Patellar Tendon- Thigh to lower leg connector under patella</td>
</tr>
<tr>
<td>• Articular Cartilage- Smooth bone end covers</td>
</tr>
<tr>
<td>• Anterior Cruciate Ligament- Thigh to lower leg connector</td>
</tr>
<tr>
<td>• Tibia- Bigger lower leg bone: shin bone</td>
</tr>
</tbody>
</table>

Figure 9 Typical Knee joint components

Typical knee joint structure is shown in Figure 9 which includes Femur and Tibia bones, patella, patellar tendon, articular cartilage and ligaments. And muscle responsible for knee joint movement is shown by Figure 10 and Figure 11. Knee joint provides a three dimensional motions [37], however, utilization of revolute joint-based planar exoskeletons will constraint the knee motion typically into knee flexion and extension.
The maximum knee flexion varies from 130-150°, which can be measured as a change of angle when subject draws their heel to back of their leg while lying on their back. Knee flexion involves hamstring muscles (e.g. - biceps femoris, semitendinosus and semimembranosus), popliteus muscle, gracilis muscle and sartorius muscles. Knee extension is the measure of straighten of the leg and its value ranges from 0-10°[38]. The muscle involved for the extensions is quadriceps femoris (e.g. - rectus femoris, vastus lateralis, and vastus medialis and vastus intermedius) and tensor fasciae latae. Internal or Medial rotation is due to semimembranosus, and
popliteus muscle while external or lateral rotation is due to the biceps femoris [39]. Anatomical visualization and change in the helical axis during motion is shown in Figure 12 (a) and (b).

![Figure 12 Patient-specific anatomical visualization of the helical axis using MRI technique [33]](image)

1.4. Motivation

Design approach for available exoskeleton is based on aligning the joint axis with the corresponding human counterpart and use of the multiple actuators for providing the desired degree of freedom [40-43]. This approach assumes that the actual axis of the movement can be accurately known. But which is somewhat different in the human joint case as the human body parts can follow complex pattern of motion. As, the center of rotation of joints changes its position during motion [44] and perfect alignment of the joints is difficult due to the variances of human leg kinematics. This imperfect alignment can create the parasitic forces in the attached points and joints [45-47]. Hence, most of the systems are incapable of actively supporting knee joint resulting poorly constrained system and unusual motion pattern which may not be ideal for real life use of the knee. It can also results pain, discomfort or long-term injuries [48]. In general, several human joints such as the hips, knee and shoulders are ball and socket joints, with the center of rotation inside the body. It is difficult for an exoskeleton to exactly match the motions of these joints using a series of external single-axis hinge points. Moreover, the compound motion of human joints
makes their alignment with an exoskeleton more difficult. And misalignment can create large stresses on the attached systems and underlying human anatomy, giving rise to the need for novel exoskeleton design strategies that permit the complex 3D motions independent of anatomical measures and landmarks. Thus, the motion of a human limb has to be studied through the state-of-the-art of motion capture and analysis techniques while making the desired clinical motions. And the captured data need to be used to determine the type/ the design of the exoskeleton accordingly.

During the rehabilitation process, an active engagement of the patient is also required to facilitate the recovery speed. Real time progress monitoring and using the status as a feedback for the process will also provide autonomy to the user and will be a good motivational aid as well.

Parallel mechanism can be used to overcome the imperfect joint alignment problem. It also has several inherent advantages over the conventional serial mechanism. It provides low effective inertia and compact design with high stiffness as the actuators can be grounded or fixed. High precision and force bandwidth are achievable due to stiff structure. The Lack of compliance of the actuation and control system in most of the existing exoskeletons is another problem for satisfactory performance [24]. Which can also be solved using EMG based control system and force signal feedback to control the mechanism as well as for recovery progress tracking.
CHAPTER 2

2. OBJECTIVE AND METHODOLOGY

2.1. Objective

This thesis aims to develop a novel task based knee rehabilitation exoskeleton which can reduce the fitting and alignment problems in an existing device. Knee injury due to stroke or other age-related disease would limit patients from accomplishing certain day-to-day living tasks. Rehabilitation with the right device and clinical practices will help to restore the lost freedom. For lower leg rehabilitation, a novel exoskeleton which can reproduce the natural motions within the human leg workspace without imposing extra load on user is required. Precise and reproducible path generation is important for getting maximum benefit from the rehabilitation process. The existing extensive research in exoskeleton design has focused on mimicking the human limb. Here, a novel, comprehensive design process to create exoskeletons based on desired tasks and human kinematics; in which, it is not necessary to know the geometry of the targeted limb, but rather to define the motion of the exoskeleton segments based on its point of attachment to the limb. This approach will help in reducing stress on the knee during rehabilitation while providing the right support and guidance. The main contributions of this work are: (1) Study of the human knee kinematics through the state-of-the-art of motion capture, (2) apply exact kinematic synthesis procedure to design a novel single degree-of-freedom (DOF) knee rehabilitation device which can reduce fitting and alignment problems, while generating the complex 3D motion of the lower leg, (3) mechanism verification through CAD simulations and prototype models, and (4) implement an assist-as-needed controller, in which the exoskeleton utilizes force and EMG feedbacks to adjust its level of support.
2.2. Methodology

This product based research starts with the needs assessment to find out the current demand and necessity of the research goal and proceeds through design and series of experiments. It includes the motion capture phase, Product development phase, controlling system design and experiment phase and final product analysis phase. The overall procedure and tasks are shown in Figure 13.

Figure 13 Methodology flow chart
After the beneficial movement pattern for faster recovery procedure has been identified, motion capture is performed using the Cortex motion capture system, while a healthy subject performs those desired movements. The data obtained from the experiment has been processed and analyzed to get the desired trajectory data.

Product development includes mechanism synthesis based on the motion capture data has been performed in Mathematica software. The solution obtained from the Mathematica is utilized for the mechanism design, followed by the detailed CAD design of the product. The product is optimized, tested and simulated for the satisfactory performance. The prototype has been developed using 3D printing Technology. Validation of the mechanism for the selected task has been performed through experiments. For testing the device, a control system has been implemented, the control system uses patient upper leg EMG signal as a control input and force sensors to get feedback responses. Finally, real-time testing of the final product to insure the feasibility, reliability and usability has been done.
CHAPTER 3

3. CHARACTERIZATION OF HUMAN KNEE KINEMATICS AND ITS WORKSPACE

This task involves collecting human motion data and identifying trajectory representation capable of reproducing the desired motions. The desired task has been traditionally specified as a set of finite precision positions that the end-effector of the kinematic chain /targeted limb connection point should pass through [49]. Recent research efforts are directed towards the use of full trajectories or regions of the space that define the workspace. Knee joint provides a three-dimensional kinematic motion of the lower leg which can be captured using state-of-the-art of motion capture. The kinematic data obtained from this stage will be used as an input to the kinematic synthesis algorithms, which in turn will generate the design equations to enable the selected linkage(s) to achieve the desired extension, flexion and pronation motion of the lower-leg.

3.1. Motion Capture System

Motion capture is the process of sensing, digitizing and recording or tracking the movement of people or objects in motion. This is mainly used in Medical applications, military operations, sports and entertainment and robotics for end effector position tracking etc. In this process, the locations of the movement of a body is tracked several number of times per second. Analysis of the skeleton movements while doing the normal activities can provide potential benefits in clinical assessments of the rehabilitation, injury prevention, and elder cares [50]. Kinematic study of these movements can be done using active [51, 52] or passive sensing methods [52, 53] of motion capture. There are basically five different methods of motion capture which are as follows.
• Prosthetic motion capture (Electromechanical Motion capture) – It uses the wearable exoskeleton with potentiometers and can give real time data with high accuracy. Data glove is one of the widely mechanical motion capture system.

• Optical fiber motion capture – it uses optical fiber sensors which measure the transmitted light by bends or twist sensors and data are less accurate.

• Electromagnetic Motion capture – It uses the central magnets attached to the subject, body with several receivers and record the movement on the computers. Its accuracy depends on the surroundings and the power of the magnet used.

• Acoustic Motion capture – it uses the audio transmitters on the subject and microphone receiver to receive the sound waves. This system accuracy is less and has very limited range.

• Optical Motion Capture – it uses multiple cameras from different viewpoints to capture the locations of the reflective markers. It is able to capture high speed data with high accuracy. This can be divided into marker based system like Cortex motion capture system and image based capture system like Microsoft Kinect skeletal tracker [53].

Due to the high accuracy and reliability, data marker based Cortex system has been utilized for the motion capture purpose. The basic approach for the cortex system is given in Figure 14 and the marker setting is shown in Figure 15. Cortex software is a complete package that can meet the most significant requirements of the motion capture applications. It can provide Marker positions and orientations; Skeletal Connectivity and Joint angle varies with the movement. The output data from this software is generated as a real time. This is a huge benefit, because this feature can make this system to get connected to lots of other 3D animation packages. Cortex software includes three major functions like the calibration of the whole captured volume. It can identify and locations of
marker in the given 3D space. It includes the tools that are required to post process the data acquired like tracking, editing and preparation of data for further use.

![Production Pipeline Diagram](image)

Figure 14 Block diagram showing motion capture system

3.2. **Experiment:**

For this project, an experiment has been conducted in the Clinical Biomechanics Instrumentation lab at WSU where the cortex motion capture system is available (Figure 16). The data has been captured for two different cases. (1) Planner case- to record the trajectory of motion of specified point in the lower leg during the extension and flexion of the knee joint, and (2) The lower leg pronation motion has been recorded for the same specific point during the rotation.
3.3. Experimental Setup

Initially, in the cortex motion capture system, the origin coordinates for the data measurement has been specified with the help of L shaped frame/band and calibration of the system has been done using 3-point reflector calibration rod. Then a stationary chair is placed in the image capturing space of the lab. To restrict the motion of the upper leg, it has been fixed to the chair using cello tape and straps. Six image capturing markers have been used for capturing the coordinate and orientation, among which three of them have been placed on the upper leg and 3 on lower leg. These markers have been placed in a specific pattern to obtain the position and orientation of the desired point.

3.4. Data Collection and processing

The extension and flexion motion followed by the pronation motion have been performed and data recorded. The collected data have been than analyzed and processed for filling the missing points in the trajectory. Smooth curves have been generated for each marker point and trials. The extension and flexion data plot are shown in Figure 17 and Figure 18. These figures show that the 2D and the 3D path of the fixed point on the lower leg during the planner movement, however the data have an out of plane component as well. It might be due to the lower limb is not actually
rotating about the fixed point or not making a “planar” motion, i.e. Knee joint is not a perfect revolute/hinge joint. Also, even though the upper leg has been fixed, there has been a slight movement as shown by the reference marker data (red color) (Figure 17) which might be due to the muscle movement during the exercise.

Figure 17 Trajectory of the fixed point during planner movement of knee

Figure 18 3D View of the trajectory during planner extension and flexion
Figure 19 shows that the motion is in the YZ plane, keeping the X axis fixed. However, there is slight movement in the x direction, which is due to the extension and flexion effect on the muscle and cannot be avoided.

The pronation trajectory of a fixed point on the lower leg has been recorded and plotted as shown in the Figure 20. During the experiment, it has been tried to maintain a circular trajectory on the XZ plane only, however, a significant component has been noticed along the out of plane (Y direction) as shown in Figure 21. This supports the claim that the knee joint axis changes during motion due to its complex structure.
Figure 20 Trajectory of the fixed point during the rotation in XZ plane

Figure 21 X4, Y4, Z4 coordinate trajectory during the rotation
Finally, both the rotation and extension-flexion experimental data have been plotted together as shown in Figure 22 to visualize the range of motion of the human knee joint and to obtain the common intersection point and utilize it for the kinematic synthesis procedure.

Figure 22 Combination of planner and rotation movement
CHAPTER 4

4. SYNTHESIS OF ROBOTIC-EXOSKELETON

This task is concerned with creating an exoskeleton robot linkage to best recreate the measured limb motions from motion capture data. Dimensional or geometric synthesis of mechanism refers to the kinematic dimensions parameter (e.g. - links length, angle, offsets etc.) determination to get the required motion behavior through graphical or analytical methods to optimize performance indices and workspace [54-57]. These Geometric parameters widely varies with the design criteria [58]. It can be done by adjusting the workspace of the kinematic structure to match exactly (exact synthesis) or approximately (approximate synthesis) the task manifold. There are several established methods for kinematics synthesis of mechanisms some well-established methods are by Sandor and Erdman [59], Hunt [60], McCarthy and Soh [61], Hartenberg and Denavit [62] and Wu at al [63]. Earliest work for motion synthesis can be found by Burmester which dealt with the precision position synthesis for five exact position for a four-bar linkage. But these finite position synthesis methods do not have the control of the trajectory and are liable to circuit defects and isometries [64]. For more than five points usually an approximate synthesis is performed through Kinematic mapping given by Ravani and Roth [61, 65, 66]. Modern kinematic mapping synthesis approach found in Bottema and Roth and McCarthy[61] uses the points in 3D space known as (image space of planar kinematics) to map planner displacements. So, in these methods, 1-DOF planner or special mechanism is shown as an intersection curve of 2 different algebraic surfaces in that Cartesian space or image space. Hence kinematic mapping provides the simpler design equations for synthesis and analysis of special mechanism.[67, 68] This approximation method is then transformed to an algebraic curve fitting problem which can utilize various approximation theories including least square problem. For the
known workspace expressions, some specific positions in the image space is able to define the workspace and even synthesize the mechanism\[69\].

For the knee rehabilitation, most of the existing exoskeletons can easily produce the flexion and extension motions, however due to the nature of its 3D motion (not on a single plane), the circular trajectory of the lower leg is difficult to be generated through the existing, and hinged joint- based planar mechanisms. Thus, in this study, a closed loop RCCR (R-Revolute, C-Cylindrical) mechanism is selected to generate the 3D circular motion of the lower-leg.

In this kinematic synthesis of a closed loop RCCR mechanism, an exact workspace synthesis approach is applied. Motion captured data is mapped into a curve in image space and cylindrical quadratic surfaces were generated. The parameterized forward kinematics equations of serial chain have been formulated and converted to the implicit equation through the elimination. Gröbner bases \[70\] approach for the specific kinematic chain have been used. This approach is relatively simple but powerful tool that uses the set of multivariate polynomial equations to solve the problem. It changes the polynomials equations to the another set of Gröbner basis which have same number of solutions but are easy to solve due to the Gröbner basis properties. The implicit description of the workspace is made to be a function of serial chain structural parameter and these parameters is related to the parameters of the given algebraic surfaces. Since only the intersection curve data which share the intersection are given, the results are not unique. The overall synthesis procedure of the exact kinematic synthesis of RCCR mechanism is shown in Figure 23, in which the intersection of the two RC serial chains is utilized to generate the desired trajectory of the knee.
Figure 23 The Exact workspace synthesis methodology applied to RCCR linkage.
4.1. Forward kinematics of closed loop RCCR linkage

RCCR linkage can be formed by joining the serial RC chain with the CR chain end effectors. The closed loop RCCR linkage is over constrained mechanism as per K-G formulae shown in equation 1. But when the revolute joint and cylindrical joint of each pair are parallel, it can move with the single degree of freedom.

\[
M = 6(n - 1) - \sum_{i=1}^{j} (6 - f_i)
\]  

(1)

The joint variable functions and geometric features can be obtained through forward kinematics of RC and CR serial chains joined at the end-effector. From Figure 24, the required parallel axis condition is \( \alpha_1 = \alpha_3 = 0 \). Also, the forward kinematics equations of RC and CR chain can be written as

\[
[D_{RC1}] = \begin{pmatrix}
    c(\theta_1 + \theta_2) & -s(\theta_1 + \theta_2)c\alpha_2 & s(\theta_1 + \theta_2)s\alpha_2 & a_2c(\theta_1 + \theta_2) + a_1c\theta_1 \\
    s(\theta_1 + \theta_2) & c(\theta_1 + \theta_2)c\alpha_2 & -c(\theta_1 + \theta_2)s\alpha_2 & a_2s(\theta_1 + \theta_2) + a_1s\theta_1 \\
    0 & s\alpha_2 & c\alpha_2 & r_2 \\
    0 & 0 & 0 & 1
\end{pmatrix}
\]  

(2)

And,
\[
[D_{RC2}] = \begin{pmatrix}
  c(\theta_3 + \theta_4) & s(\theta_3 + \theta_4) & 0 & -a_2 c \theta_4 - a_4 \\
-s(\theta_3 + \theta_4) c a_4 & c(\theta_3 + \theta_4) c a_4 & s a_4 & -r_3 s a_4 + a_3 s \theta_4 c a_4 \\
s(\theta_3 + \theta_4) s a_4 & -c(\theta_3 + \theta_4) s a_4 c a_4 & c a_4 & -r_3 c a_4 - a_3 s \theta_4 s a_4 \\
0 & 0 & 0 & 1
\end{pmatrix}
\]  

(Where c is cos and s is sin function)

Equating these two transformation equations, angular relations and geometrical constraints can be derived.

In order to characterize the workspace of the parallel RC chain, implicitization is performed to eliminate the joint variables \( \theta_1 \) and \( r_2 \). The elimination yields a quadratic surface expression.

\[
Q (X, Y, Z) = (S_{1Y}^2 + S_{1Z}^2) X^2 + (S_{1X}^2 + S_{1Z}^2) Y^2 + (S_{1X}^2 + S_{1Y}^2) Z^2 - 2S_{1X} S_{1Y} X Y - 2S_{1X} S_{1Z} X Z - 2S_{1Y} S_{1Z} Y Z + C_{21X} X + C_{21Y} Y + C_{21Z} Z = 0
\]  

(4)

Where, \((X, Y, Z) = A \text{ point of } R^3 \text{ space of relative translation}

\(S_1 = (S_{1X}, S_{1Y}, S_{1Z}) = \text{direction of both the R and C joints, and } C_{21} = C_2 - C_1 = (C_{21X}, C_{21Y}, C_{21Z}) = \text{vector along common normal between the joints.}

The surface of circular cylinder with radius \( R = \sqrt{C_{21} \cdot C_{21}} \) with zero relative displacement and passing through origin is obtained. And hence workspace for RCCR link is the combination of two circular cylinders resulting constant orientation quartic curve. The Points obtained from the motion capture for the lower-leg motion have been used to solve this quartic equation to exactly fit the trajectory. Given an initial point \( P_1 \) (which is used as reference configuration), relative displacements of the RC-CR chain will move this point to the rest of task points \( P_2, P_3, P_4, \ldots P_n \).
The action of the chain on this point can be calculated using one of the conjugations in the Clifford algebra as shown in Equation 5.

\[ P_i = \bar{D} \hat{P}_1 \bar{D}^+, \quad i = 2, 3, ..., n. \]  

(5)

Where \( \hat{P} = 1 + \varepsilon P_1 \) = dual quaternion expression of point P1.

Let the value of \( \theta_1 \) and \( r_2 \) be independent in parallel RC chain, this gives 3(n-1) design equations, structural variable \( S_1 \), joint variable \( \theta_1 \) and \( r_2 \) and \( C_{21} = C_2 - C_1 \) for each joint and 4+2(n-1) total unknowns. For doing the exact path point synthesis, maximum of 5 points (n=5) can be defined. Shape of the circular cylinder and parallel RC chain motion path is defined using four relative translations positions \( P_2, P_3, P_4, \) and \( P_5 \). Hence design equation has six quadratic equation and six unknowns.

\[ Q(P_i) = 0, \quad i = 2, 3, 4, 5; \]

\[ S_1.S_1 = 1, \quad S_1.C_{21} = 0, \]  

(6)

4.2. Application of the Kinematics synthesis into the generation of lower leg trajectory

Figure 25 shows the selected path for both the planner and rotation data for the synthesis of the mechanism. In this study, however the 3D circular motion of the lower-leg is considered in the design, as the planar motion can be generated easily through the existing planner mechanisms. Data points, sampled from the selected path (Table 2) have been utilized for the mechanism synthesis.
Table 2

Task points used for the RCCR chain simulation

<table>
<thead>
<tr>
<th>Points</th>
<th>Coordinates</th>
</tr>
</thead>
<tbody>
<tr>
<td>P1</td>
<td>58.4111, 345.165, 258.234</td>
</tr>
<tr>
<td>P2</td>
<td>34.737, 351.698, 201.175</td>
</tr>
<tr>
<td>P3</td>
<td>-6.34155, 370.433, 188.565</td>
</tr>
<tr>
<td>P4</td>
<td>-31.6289, 383.232, 264.294</td>
</tr>
<tr>
<td>P5</td>
<td>18.7943, 379.871, 313.63</td>
</tr>
</tbody>
</table>
The outlined exact synthesis procedure provides sets of solutions, any combination of these solutions will provide the required RCCR mechanism, i.e. the intersection of these two solutions generates the workspace of the RCCR linkage, which is the desired lower-leg trajectory. One of the selected solution is shown in table 3. The intersection of the two serial chains of this solution is shown in Figure 26, the axes direction and positions are shown in table 3. The solution is then assembled and used to create RCCR linkage. The linkage is then validated for its accuracy in generating the desired trajectory.

<table>
<thead>
<tr>
<th>Axis</th>
<th>Si</th>
<th>C2 – C1</th>
</tr>
</thead>
<tbody>
<tr>
<td>S1</td>
<td>-2.85861, 252.316, 384.131</td>
<td>-72.9612, 220.26, -5.0004</td>
</tr>
<tr>
<td>S2</td>
<td>19.0904, 316.512, -29.5126</td>
<td>70.6081, 340.069, 256.456</td>
</tr>
<tr>
<td>S3</td>
<td>46.2141, 350.26, 260.011</td>
<td>40.9261, 328.67, 285.614</td>
</tr>
<tr>
<td>S4</td>
<td>-5.24389, 344.451, 289.385</td>
<td>-25.4944, 261.772, 387.43</td>
</tr>
</tbody>
</table>

Figure 26 Solution from the synthesis for the circular motion alone
To validate the synthesis approach, the output path of the mechanism should match the desired clinical trajectory obtained from a motion capture system. The selected data path obtained from the motion capture system has been plotted. Similarly, the end effector path of the synthesized mechanism has been generated for comparisons. These both trajectory was plotted together as shown in Figure 27. The trajectory obtained through the mechanism and the desired trajectory obtained through the motion capture are identical, considering the original trajectory from the motion capture is a representative of several paths with few offsets.

Figure 27 CAD model of the selected solution and trajectory (blue) generated by mechanism with the motion capture data (red)

CHAPTER 5

5. PRODUCT DESIGN AND TESTING

5.1. Design of the selected Mechanism in a CAD Environment

The selected mechanism has been designed in CATIA. The design of this exoskeleton device includes five major parts, namely base support, motor gearbox assembly, active crank link, passive follower link, coupler link (end-effector). The coupler link and the end-effector are fixed together by a pin. The design was done in several iterations for the better performance and product reliability. The first design of the device, shown in Figure 28, had a good accessibility for ease of
utilization and avoiding interference with the user, however, had the force distribution problem, also it also creates a big stress concentration on the part.

Figure 28 First cad design of the synthesized mechanism

This design was then modified to prevent the possible force distribution problem. Initially, the end effector profile was changed to curved surfaces and integrated with coupler link to form single part. This new design showed the improved performance and withstands the grater point force in the tip of the end effector. The modified design is shown in Figure 29.
5.2. Detailed description of the parts

5.2.1. Base Support

The base provides the stability and support to the mechanism. It has provision to accommodate NEMA motor in one side, which is shown by the hole and is the active end of the mechanism. Another end provides a guide for the rotational joint and is attached to the passive connective link. These both end is specially designed considering the synthesized axis, hence orientation is less irregular. Figure 30 shows the base design of the mechanism.
5.2.2. Motor Gearbox Assembly

The standard design of NEMA 23 stepper motor with the gearbox assembly, 23HS22-2804S-PG47 has been utilized in the mechanism which have 56mm motor length, 72 mm of gearbox length and 30 mm shaft length. NEMA 23 motor is shown in the Figure 31 [71].

Figure 31 NEMA 23 motor with PG47 gear assembly
5.2.3. Active Crank Link

This link is attached to the shaft of the NEMA motor in one end and with the coupler link in another end. The axis of both holes is designed in such a way that it matches the synthesized special axis, so the holes are not perpendicular to the plane. The design of this link is shown in Figure 32.

![Figure 32 Active connective link](image)

5.2.4. Coupler Link

This link connects the active and passive connective links. Any point in this link can produce the desired clinical trajectory and the end-effector link is fixed to this link. The design of the coupler link is shown in Figure 33.
5.2.5. End-Effector

End effector connects the mechanism with the human lower leg. In the first iteration, the end effector part is integrated with the coupler link to form a single part to get better mechanical characteristics. The typical design is given in Figure 34.
5.2.6. Passive follower link

This link connects the coupler link with the base frame. It is little longer than the active crank link. The design is shown in Figure 35.
5.3. Prototyping of the designed Mechanism

Prototyping of the design was done for testing and application of the control system. Each part of the device was printed using the 3D printing technology and the parts were assembled to the final model. Figure 36 shows the initial prototype of final assembled rehabilitation device.
The choice of the appropriate material for prototyping was one of the major issues for insuring the mechanical rigidity and product viability.

5.3.1. Material Selection

For this robotic exoskeleton prototype, Polylactic Acid (PLA) has been utilized because of its high stiffness, low chance of failure, and low strength to weight ratio. The exoskeleton will
have a low weight which reduces the energy required to carry it during the operation. PLA will also help combat against wear and tear of the mechanism thereby increasing the lifespan of the robot and saving money in the long run. One of the major benefit of the PLA material is that it is bioplastic material and it naturally degrades in very short period of time in external environment reducing the robot’s impact on the environment. But despite being degradable in external environment, it is extremely robust in case of the normal applications. The cost of PLA is another advantage and, it will reduce the amount of power used by the robot hence little low power actuator can be used. As the 3D printing technology has been used for the prototyping purpose, PLA material in the form of PLA filament (in Figure 37) has been used. The properties of the PLA materials are shown by table 4.

Table 4
PLA material Properties

<table>
<thead>
<tr>
<th>SN</th>
<th>Properties</th>
<th>Values</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Technical Name</td>
<td>Polylactic Acid (PLA)</td>
</tr>
<tr>
<td>2</td>
<td>Chemical Formula</td>
<td>(C3H4O2)n</td>
</tr>
<tr>
<td>3</td>
<td>Melt Temperature</td>
<td>157 - 170 °C (315 - 338 °F)</td>
</tr>
<tr>
<td>4</td>
<td>Typical Injection Molding</td>
<td>178 - 240 °C (353 - 464 °F)</td>
</tr>
<tr>
<td>6</td>
<td>Tensile Strength</td>
<td>61 - 66 MPa (8840 - 9500 PSI)</td>
</tr>
<tr>
<td>7</td>
<td>Flexural Strength</td>
<td>48 - 110 MPa (6,950 - 16,000 PSI)</td>
</tr>
<tr>
<td>8</td>
<td>Specific Gravity</td>
<td>1.24</td>
</tr>
<tr>
<td>9</td>
<td>elastic moduli</td>
<td>3368 MP</td>
</tr>
</tbody>
</table>
5.4. **Detailed Analysis of Mechanism**

The designed mechanism was analyzed to figure out the kinematic behaviors and the end effector response with the dynamic loading using stress strain analysis. DMU Kinematics feature in CATIA V5 was used for the analysis of the mechanism.

5.4.1. **Simulation**

The assembled mechanism has been simulated in the CATIA using DMU Kinematics tool to verify the desired trajectory of the mechanism. For the simulation, the base has been fixed, to which the motor gearbox assembly has been attached and joined together. Revolute joint between the motor shaft and the active crank link has been created and made as driving joint. Another end is connected to coupler link using cylindrical joint. Also for the passive connecting link, one end is connected to base support with revolute joint and another to coupler link using cylindrical link. Hence the mechanism behaves as a RCCR joint mechanism. The end-effector has been then fixed with the coupler link. The crank link has been actuated to rotate 360 degrees to get the desired trajectory of the mechanism. Figure 38 and Figure 39 shows the trajectory of the mechanism.

![Figure 38 Mechanism attached with the lower leg and simulation trajectory](image-url)
5.4.2. Trajectory Analysis

The end-effector path of the designed mechanism has been traced to validate with the motion capture (desired) trajectory as shown in Figure 40. The trajectory traced in blue color was obtained from tracing the point in the mechanism end-effector whereas the trajectory in the red color was the motion captured data. It shows that the motion capture path is a bit irregular in shape, which is due to the human errors factors and nature of the human knee kinematics, whereas mechanism output is smooth and have regular path. But in overall, the synthesized mechanism has successfully mimicked the desired trajectory with minimum errors.
5.4.3. Rotation of the passive connective link

Unlike driving link, the rotation motion of the passive connective link is constrained to certain degree only. This rotation can be observed using Figure 41. It shows that the maximum rotation of the link is 43.19 degrees.

![Figure 41 Total angular rotation of the passive connective link](image)

5.4.4. Linear displacement of end-effector

Figure 42 shows that the maximum linear displacement of the end effector (131.53 mm). The span is small due to the original marker placement during the motion capture, the markers have been placed in the upper half of the lower leg giving smaller diameter path.

![Figure 42 End effector linear displacement](image)
5.4.5. **Stress Strain analysis**

To check for the mechanical design integrity of the designed exoskeleton, stress analysis has been performed. Load has been applied at the tip of the end effector. The applied load is equal to the load of the foot and load of the shank of a person which are 1.43% and 4.75% of the body weight respectively. Taking the standard weight 81kg (or 178.5 lbs) of the man of height 170 cm (5 ft 7 in), the foot weight is 1154 gm and shank weight as 3848 gm. Hence the total weight to be applied at the mechanism is 5002 gm [72]. But for the safety provision, considering the factor of safety of 2, total weight of 10004 gm or 98.11N is applied. Figure 43 shows that even at factor of safety 2, the maximum von Mises stress is 969 MP which is less than the stress that the material can handle so the design is safe.

![Figure 43 Displacement vector and von Mises stress on the end effectors](image)
CHAPTER 6

6. ACTUATION SYSTEM

The choice of an appropriate actuator is pivotal in the design of exoskeleton for obtaining different functionality as those deliver the mechanical force that is needed to overcome the machine’s weight while providing the required support to the user. Some commonly used actuators include hydraulic actuators, pneumatic actuators and electric motors, etc. Hydraulic actuators use pressurized oil to move the exoskeleton joints. This actuator type has very large potential force, torque and accuracy but they are generally heavy, noisy, require a complex apparatus to work (compressor, cooling system, etc.) and have a potential to leak [73]. Pneumatic actuators use pressurized air to produce motion. Comparing with hydraulic actuators, they are lighter and cheaper. Additionally, they provide a clean, non-flammable material, on the other hand, they also require a compressor or a rechargeable pressured tank also it is unpredictable for precise motion control [2, 73]. Electric motors are the most widely used ones due to their simplicity to install, lightweight, small, low noise, and offer a clean actuation system. In addition, electrical servomotors are efficient and provide a higher power-density, and by using high-gauss permanent magnets and step-down gearing, they can provide high torque and fast and precise motion but they may require extra energy for maintaining a steady position. [73] for example MS-02 Power Loader made by Korean Daewoo Shipbuilding and Marine Engineering was used to increase the strength of the workers and used electrical motors as well as The ReWalk exoskeleton [2]

Ionic polymer metal composites (IPMCs) are electro-active polymers with smart sensing and actuation capacity inherited within them. As electrode, Nafion, a thin ion exchange membrane chemically plated in some highly conductive metal is used. The actuation occurs as a bending effect due to movement of hydrated cation under applied electric field. IPMCs are very useful in
biomimetic robots and biomechanical devices due to its resilience, softness, deformation producing capability and bio-compatibility[74]. But these materials are not much studied for the real-time control system and are not capable of lifting large loads. Shape memory alloys actuators are made from titanium-nickel (TiNi) and can generate small displacements but cannot be widely utilized because of the difficulties to manufacture and are quite expensive as well [75, 76]. Elbow exoskeleton actuated with shape memory alloy is proved to be light weight, simple structure and less noisy [77].

In this project, NEMA 23 stepper motor of 56mm body with 2.8A rated current having integrated a planetary gearbox of 46.656:1 gear ratio has been used for the mechanism. It provides high torque and desired low speed. The Electrical, gearbox and physical specifications of used motor are as listed in Table 4.

Table 5
NEMA 23 specifications [71]

<table>
<thead>
<tr>
<th>Electrical Specification</th>
<th>Gearbox Specifications</th>
<th>Physical Specifications</th>
</tr>
</thead>
<tbody>
<tr>
<td>Manufacturer Part Number: 23HS22-2804S-PG47</td>
<td>Gearbox Type: Planetary</td>
<td>Frame Size: 60 x 60mm</td>
</tr>
<tr>
<td>Motor Type: Bipolar Stepper</td>
<td>Gear Ratio: 46.656:1</td>
<td>Motor Length: 56mm</td>
</tr>
<tr>
<td>Step Angle: 0.039 deg</td>
<td>Efficiency: 73%</td>
<td>Gearbox Length: 72mm</td>
</tr>
<tr>
<td>Holding Torque without Gearbox: 1.25Nm(177.01oz.in)</td>
<td>Backlash at No-load: &lt;=1.5 deg</td>
<td>Shaft Diameter: Φ12mm</td>
</tr>
<tr>
<td>Rated Current/phase: 2.8A</td>
<td>Max.Permissible Torque: 40Nm(5664oz.in)</td>
<td>Shaft Length: 30mm</td>
</tr>
<tr>
<td>Phase Resistance: 0.9ohms</td>
<td>Moment Permissible Torque: 60Nm(8497oz.in)</td>
<td>Key-way length: 20mm</td>
</tr>
<tr>
<td>Voltage: 2.6V</td>
<td>Shaft Maximum Axial Load: 100N</td>
<td>Key-way width: 4mm</td>
</tr>
<tr>
<td>Inductance: 2.5mH ± 20%(1KHz)</td>
<td>Shaft Maximum Radial Load: 200N</td>
<td>Number of Leads: 4</td>
</tr>
</tbody>
</table>
6.1. **Motor and gearbox dynamic equation**

Dynamic equations for the motor and gearbox can be derived for getting the output characteristics which are given as:

- moment of inertia of the rotor \((J)\)
- damping ratio of the mechanical system \((B)\)
- electromotive force constant \((K=K_e=K_t)\)
- electric resistance \((R)\)
- electric inductance \((L)\)
- input \((V)\): Source Voltage
- output \((\theta)\): position of shaft
\[
\dot{\theta} + B\dot{\theta} = Ki
\]
\[
L \frac{di}{dt} + Ri = V - K\dot{\theta}
\]

In order to control the position, the transfer function will be

\[
s(js + B)\theta(s) = Ki(s)
\]
\[
(Ls + R)i(s) = V - Ks\theta(s)
\]

Looking at the position, and \( \theta \) being the output. We can obtain the position by integrating \( \text{Theta Dot} \), therefore we just need to divide the transfer function by \( s \).

\[
\frac{\theta}{V} = \frac{K}{s(js + b)(Ls + R) + K^2}
\]
Pioneering breakthrough in the controlling system has brought the precision, accuracy and smartness in the control system. Sensor is very critical to do precision activities and improve controllability. Controlling of the robot using human motion intension is very important and challenging task. To fully utilize and benefit from exoskeletons, proper control method selection is very crucial [78]. Study on biological signals like Electromyography (EMG), Mechanomyogram (MMG), Electrooculography (EOG), Electrocorticogram (EcoG) and Electroencephalography (EEG) is being done as a control input for different controllers [79]. Every human activity is related with the movement of the muscle creating EMG and other signals which can accurately define different motions or intention of motion or human biomechanics [80]. Among these biological signals, EMG has given promising results for controlling robots with correct human motion intension interpretation [81], also can be analyzed to detect medical abnormalities, level of injuries and recovery [82, 83]. Hence EMG based control systems are important for the control of high level rehabilitation system. Hybrid control system uses control input from two or more systems and is able to combine the advantages of different system used but is more complex [84]. Future controlling prospects include enabling the user to control the exoskeleton by thought. Currently, all motion is activated by detecting the motion of the user. There is currently a project working towards meeting this goal. The Mindwalker project, funded by the European Union, is a lower-body exoskeleton that is controlled by the user’s brain signals using electroencephalographic sensors on the user’s head as well as electromyographic sensors on the shoulders.
In this project, EMG based controller along with force sensors and feedbacks are employed for the implementation of assist-as-needed controller. In which, the exoskeleton will provide support according to the user’s strength, i.e. the exoskeleton provides 100% support at the beginning of the rehabilitation and it reduces its support as the user recovers some strength. In full recovery, the mechanism acts like an exercise machine, in which the user requires more energy to pull/carry the exoskeleton.

7.1. Control Strategies

![Hierarchical Controller](image)

Figure 46 lower limb prosthesis/orthoses generalized control framework [85]

As depicted in Figure 46, the controller for the P/O controlling in any exoskeleton can be generalized to be divided into three levels. The high-level controller is accountable for perceiving the user’s locomotive intent, which is determined based on signals from the user (MGE), environment, and the device through a combination of activity mode detection and direct volitional control. After that the processed data has been all passed to the mid-level controller, which converts the user’s motion aims to a desired output state for the device. In fact, this stage may consist of a combination of joint positions, velocities, and torques. Finally, the commands will be
given to the low-level controller, which represents the device-specific control loop that executes the desired movement using controllers [86].

7.2. Electromyography

Electromyography (EMG) is one of the electrodiagnostic medicine techniques utilized for recording and evaluating the electrical signal of skeletal muscles activity to nervous stimulation [82, 87]. EMG signals are highly fluctuating, noisy, time varying, and muscle activity for the same activity in different persons may differ so the use of EMG signals for pattern recognition is not an easy task [88]. Electromyography signal-controlled exoskeletons are the most advanced generations which provide greater control for movement, easy to use but they are heavy, have battery problems and the most, their constant use weakens the EMG signals which lowers the performance and increases the fatigues. Loss of muscles because of amputation affects controllability due to disturbance or unavailability of required signals [84]. Using multiple surface EMG sensors for controlling increases difficulties in electrode shifting for amputees, so it makes them uncomfortable also it is not practical. Controllability, flexibility, and reliability can be increased with limited numbers of sensors [89]. Patient evidence shows the limited uptake or even refusal to wear myoelectric systems saying that they are heavy, lack control, clumsy and ultimately not able to increase quality of life. For increasing the quality of life, focus should lay on weight, strength, functionality, comfortable socket fit and be lifelike [90].

As each activity is related to the related muscle activation pattern, specific movement patterns can be identified by different pattern recognition schemes like classification and feature extraction of the EMG signals from involved muscle [91]. Autoregressive coefficients (AR), Fast Fourier Transform (FFT) analysis of frequency and power, amplitude and wavelet analysis are some feature extraction techniques whereas artificial intelligence techniques or statistical
techniques like Support vector machine (SVM) are popular classification techniques. SVM is more suitable for regression problems and classification as SVM restrict its local and global solution from multiple local minima problem unlike artificial neural network. also, input space dimensionality does not have any effect on the final complexity of model [92]. EMG signal control methodologies can be divided in two groups, Pattern recognition based control[93] and non-pattern recognition based control system [94]. EMG signal based controlling procedure include EMG signal detection from targeted muscle, processing of the signal, classification for the desired motion or intension (this is not done in non-pattern recognition based control method) and application to control system. Detection of the signal can be done using invasive (eg Surface EMG) or non-invasive way (e.g. - intramuscular EMG). Surface EMG are little weaker and related with noises, but invasive techniques are riskier. Sometimes, SEMG are not suitable for study. For example, with the increase in the level of amputation, the EMG control signal becomes weaker and ultimately vanish. So, sEMG signals controllers only are not efficient to classify accurate motions. Targeted Muscle Re-innervation (TMR) invasive surgical procedure is able to address this problem with good accuracy but involves risk with it [84].

EMG Signal are associated with the noises and variances so are needed to be processed for continuous and reliable data. Basically, EMG signal have transient state (signal while muscle goes from rest to the voluntary state) and steady state (signal during constant contraction period). Transient state signal have large variation and errors hence steady state signal is used for the signal analysis [81]. Processing method differs in pattern recognition and non-pattern recognition technique.

Intension of motion and motion pattern can be recognized with the classification of the signal. Factors like fatigue, perspiration and electrode position changes the EMG pattern. Hence,
influence of those factors need to be neglected from the base pattern and can be done through the classification. And finally, the classified signal is given as an input to the controller to achieve intended motion pattern.

7.2.1. EMG system and its implementation

EMG is the measurement of electrical stimulation created by muscle cells due to the activity of the muscle which is in phase lag with the muscle activity by 100 ms. A mathematical model can be made to define the relationship between EMG, forces and the moments of the limbs. Thus, the robot can perform the required action to reinforce the body motion.

![Diagram of EMG signal acquisition and process]

Figure 47 EMG signal acquisition and process

EMG signal differs from person to person and in different condition. Basically, it is just the replica of the activity and force combined. The person with disability, weakness or injury produce significantly different pattern and strength signal than to normal person [95]

7.2.2. EMG Data capture and implementation

sEMG signals have been obtained using DELSYS TRINGOTM EMG System. This system is equipped with the wireless EMG sensors with the range of 40m and has built-in triaxial accelerometer and a base Station which has recharging cradles for sensors and full trigger capability. Hybrid and mini head sensors are available with the signal resolution of 168nV/bit and sampling rate of 2000-4000 samples per sec. Signals generated from the sensors are in the bandwidth of 20-450 HZ with 16-bit signal resolution. The inbuilt accelerometer is in the
sensitivity of ± 1.5g or ±6g and can measures 148.1/296.3 samples per sec. The base Station provides the communication and power feedback through LEDs and has analog output range of ± 5V. This wireless system is capable of transmitting data from the 16 EMG sensors to the EMG works Acquisition and Analysis software and generates 16 distinct EMG signals and 48 accelerometer positions with the delay of 48ms for the EMG signals and 9600ms for the accelerometer positions. The generated signals can be analyzed using EMG Works Analysis which can use different calculation script for better comparison.

Advancer technologies myoware sensor were used for the EMG signal acquisition during the experiment as it is cheap and is easy to use, also it can be connected to the Arduino board
directly for control input. This sensor provides the output raw EMG signal in the range of 0-5v depending on the muscle activity hence; the signal doesn’t need to be amplified.

7.2.3. EMG signal sensor Installation

Selecting the proper muscle related to the motion is essential to detect the EMG signal that will be used as the input for the control system. As per the medical expert the following muscles are responsible for the Knee joint flexion.

- Biceps femoris
- Gastrocnemius
- Gracilis
- Popliteus
- Sartorius
- Semimembranosus
- Semitendinosus

Figure 50 Knee joint Muscles
However as per [96] the best way to collect data is from the following muscles gluteus maximus (GMAX), vastus medialis (VM), rectus femoris (RF), semitendinosus (ST), and gastrocnemius (GAS)[97]. Based on the measured EMG data, muscle activation, $A_{ch}$, will be categorized into two conditions: “activated” and “inactivated” by sing Single-threshold method.

$$
A_{ch}(k) = \begin{cases} 
1 & \text{(activated)}, \\
0 & \text{(in activated)}, \\
\text{Otherwise} 
\end{cases} \quad \text{if } \chi_{ch}(k) \geq Z_{ch} \tag{9}
$$

where $\chi_{ch}(k)$ is the amplified, band-pass filtered, and full-wave rectified EMG signal of each channel at a discrete time instant $k$. $Z_{ch}$ is the threshold value that equals to the mean plus three standard deviations (Mean + 3SD) of $\chi_{ch}$ when the muscle is relaxed [98].

### 7.3. Controller design algorithm

This controller design makes use of Arduino for real time controlling of the exoskeleton. The revolute joint is actuated with the NEMA 23 stepper motor. The real time sEMG signals were acquired from Rectus Femoris (RF) muscle which was filtered and rectified using MATLAB interphase and threshold value of the signal was applied to drive the motor. The basic diagram for motion control of the exoskeleton device is shown in Figure 51.
During this rehabilitation progress, recovery of the patient is monitored through the force sensors embedded in the exoskeleton. This feedback signal is obtained through the force sensor installed in exoskeleton which senses the amount of force exerted by the exoskeleton to the leg. At the initial condition, injured leg have weak power hence exoskeleton needs to apply large amount of force. As the patient recovers, the force requirement eventually decreases with time so as pressure applied by the exoskeleton to leg. When the patient is fully recovered, all the effort to create the desired trajectory is applied by patient himself and actuator uses no power, which will provide zero reading in the force sensor indicating the full recovery. The basic block diagram for recovery tracking is shown in Figure 52.
7.4. **Control system implementation and experiment**

![Figure 53 Experimental setup for control system implementation.](image)

The experiment was conducted to control the lower leg motion through exoskeleton using EMG signal as an input to the Arduino based controller. Advancer technologies myoware sensor has been utilized to get the real time EMG signal from Rectus Femoris muscle to the Arduino. MBC12101 motor driver has been used to provide the Arduino output to the NEMA 23 motor.
Experimental setup is shown in Figure 53 in which all the necessary connections were made as shown in Figure 54.

![Circuit diagram for controlling exoskeleton](image)

Figure 54 Circuit diagram for controlling exoskeleton

After all the setups have been done, controlling algorithm for the assist as needed controlling strategy was created. This controlling strategy uses both the EMG and FSR sensors to control the intention of motion. Initially sEMG output value was analyzed to know the threshold value for motion intension. The mechanism has been made to operate autonomously in a specified direction based on an EMG signal, i.e. if the value of the EMG signal is greater than the threshold value, it will move in one direction, and if it is lower, it will move in opposite direction. So, using EMG signal, direction control of the motion can be done. Force sensor is utilized to constraint the motion of the exoskeleton, if the force output reaches and tend to cross the maximum specified value, the mechanism will be forced to stop.

This mechanism analyses both the EMG and FSR signal for the power control. In the initial phase of recovery, the power required for mechanism to follow the desired trajectory is maximum.
This leads to lower recovery percentage output and vice-versa. So, the amount of power input is made directly proportional to the function of the EMG signal output value and inversely proportional to the FSR sensor output value.

7.4.1. Results

The real-time control algorithm has been successfully implemented to control the motion of the end effector. When there is no EMG signal detected, mechanism will not be functional (ie when the EMG sensor was not attached to the muscle or no signal is detected). But as soon as the sensor was attached, the mechanism started moving in clockwise direction, and when the subject tried to rotate the leg (ie motion intension) the EMG value crossed the specific threshold value and rotate anticlockwise. At the same time, when the force on the FSR sensor is greater than the specified weight, the FSR sensor voltage output shows the maximum value, which send a command to stop the mechanism. The signal captured during the rotational motion of the knee is shown in Figure 55.

![Figure 55 EMG signal from Rectus Femoris muscle during rotational motion](image)

Figure 55 EMG signal from Rectus Femoris muscle during rotational motion
8. RECOVERY TRACKING

Recovery tracking is one of the unique features of this exoskeleton device which makes it more automated and user friendly. The tracking of recovery has been done using the force sensor signal. The sensor applied at the contact point of the end effector and the lower leg provides the amount of force applied by the exoskeleton on the human leg. This information is analyzed and utilized for the recovery tracking process.

8.1. Force sensing resistors

Force sensing resistors are flexible device based on polymer thick film technology. Its resistance decreases with the increase in the amount of applied force. The typical force resistance characteristic is shown in Figure 56. However, the characteristics slightly differ in case of different FSR as the change in design and thickness changes the sensitivity. A true force sensor is capable of producing constant reading for a constant force. These can be used for a longer duration and also on elevated temperatures [99].

![Figure 56 Typical FSR force resistance characteristics][99]
Figure 57 Typical force sensing resistor design

Figure 57 shows typical design of polymer film FSR which consists of 3 different layers. In the prototype, the FSR is applied at the point of contact between the end-effector of a parallel exoskeleton robot and lower leg. Hence, it measures the force applied by the end-effector to the leg to move it in the predefined trajectory. This obtained force signal is given as input to the controller as a feedback signal. This signal is used for two basic purposes, recovery tracking and end-effector power control.

Initially, the amount of force required to move the leg is more, which gradually decreases as the recovery occurs and less force is needed to do the same task. So, the amount of force applied can be compared to the degree of recovery. Lesser the force applied by the mechanism on the leg, higher is the recovery and vice versa. Also, if the leg is being able to apply some force through its own muscles, the amount of help needed from the exoskeleton is reduced which can also be controlled by force sensor data reading.
8.2. Recovery Progress Tracking Methodology

The technique for recovery tracking is based on the amount of force applied by the exoskeleton to the human leg. In the beginning, when the patient is unable to move the leg, the total force provided by the device is equal to or greater than the weight of the lower leg. Total force can be greater than the weight itself because of the muscle constraints (due to swelling and related effects). Ideally, for a healthy person, the total force exerted by the exoskeleton to do the task is assumed to be zero. Hence, this difference of the force is converted to percentage scale and termed as the recovery scale.

Let, the force applied by the weight of the lower leg = \( W \) (6.18\% of total body Weight). Because of the recovery, the patient is able to put some effort on his/her own. At that moment, the force applied by the exoskeleton to the leg is less than the total force \( W \), suppose \( F \). Decrement of the effort put on the leg by the exoskeleton is equal to the difference between the maximum force and the actual force applied by the exoskeleton \((W-F)\). Hence, total recovery can be obtained by using eqn. 12.

\[
Recovery = \frac{W - F}{W} \times 100 \%
\]  

When the applied force by the exoskeleton is equal to the total weight \((W)\), (i.e. \(F=W\)), the leg is fully constrained and needs recovery. The recovery scale is Zero. But, when the leg starts healing/recovering, the amount of force from the exoskeleton decreases and an improved/increased recovery percentage scale will be observed. When the patient is fully healed (i.e. \(F=0\)) the recovery scale becomes 100\% and the patient does not need any further assistance.
This principle can be applied using the force signal through voltage analysis. The output voltage from the FSR circuit can be given in eqn. 13.

\[ V_{Out} = \frac{R_M V}{R_M V + R_{FSR}} \]  

(11)

Where, \( R_M \) is measuring resistance through which the sensitivity and current in the circuit can be controlled. \( R_{FSR} \) is the resistance in force sensor resistor which varies with the application of force. The amount of force applied is directly proportional to the voltage output. The higher is the force applied by the exoskeleton, output voltage is higher and as the amount of force decreases, the signal gets weaker and vanished when force applied becomes zero.

For experimentation in this device, the experimental data in Figure 58 for the RM value of 10K was chosen and the data were fitted in the curve using the curve fitting method using in a power curve. The obtained curve and the curve equation based on the output, the applied force can be found using eqn. 14.
Figure 59 Curve fitting in power curve

\[ F = 4.970873 \cdot V_{out}^{4.3^{12619}} \]  

(12)

8.3. Implementation of recovery tracking algorithm

A MATLAB program, algorithm has been created using the equation 12, 13 and 14. To test the model, an experiment has been conducted to figure out the response of the output voltage to the recovery percentage and the force applied. The Arduino-based interfacing has been created between the MATLAB and the FSR to run the experiment. The experimental setup for the task is shown in Figure 60.
8.3.1. Results

Voltage output and Recovery percentage scale has been plotted with respect to time to find out the model effectiveness. Figure 61 shows that when the voltage output is maximum, the recovery percentage scale becomes minimum and vice versa. If the applied force is more than the lower leg weight (due to swelling or any other restraints), the recovery scale goes to negative value, which indicates that there is no recovery at all.
Again, the output voltage data were plotted against the applied force to compare the results with the standard data plotted in Figure 62, with data shown in Figure 58 FSR voltage divider circuit (left) and different output voltage with force applied (right). The result is similar and hence, proves the assumption to successfully track the recovery percentage.
9. CONCLUSIONS AND FUTURE WORKS

9.1. Discussions

The design presented in this thesis has successfully generated the desired knee rehabilitation trajectory, while providing the necessary support. The mechanism does not impose any extra load on the user as it is not aligned to any of the human joint position and is not also placed on top of the patient’s body. Due to the added assist-as-needed controller feature, the device can be utilized in the home-based rehabilitation exercise without or with minimal assistance from physical therapist. Hence, the system will help in reducing the labor costs involved. Provision of recovery feedback is expected to make the patient more motivated towards the rehabilitation activities with high morale of recovery. Some of the major advantages of the designed rehabilitation device are summarized as

- Eliminates the fitting and alignment problems as it is not designed based on an exoskeleton-human joint alignment approaches.
- There will not be an extra load on the user, as it is mounted off the human body.
- Smart controlling technique with recovery tracking makes it more user friendly.
- Use of 3D printing using PLA material makes it cost effective and light weight while retaining the rigidity of the system.

While having the above advantages, this mechanism also has limitations as it only provides a specific trajectory. The patient will not have options to practice other exercises.

9.2. Conclusions

In this study, the human knee kinematics have been studied through the state-of-the-art of motion capture. The result has shown that the lower-leg motion is not on a single plane but complex
3D. Mimicking the knee joint in a single plane of motion is not accurate and is associated with many limitations hence; it needs a different approach of rehabilitation exoskeleton device design. To solve this, the exact kinematic synthesis procedure with single degree-of-freedom (DOF) parallel mechanism has been proposed for knee rehabilitation. The device able to reduce fitting and alignment problems, while generating the complex 3D motion of the lower leg. It also ensures smoothness of motion between task positions. Through the outlined design approach, human knee joint does not need to align to its counterpart mechanism joint, as the joints of the mechanism are oriented/located independent of the anatomical landmarks of the knee joint. The designed mechanism has been verified through CAD simulations and prototyping. The real time sEMG based control with the force sensor signal feedback has been implemented. This feature provides the patient to accurately control the movement and track the recovery at the same time. This additional control and sensing features the Knee rehabilitation device user friendly and reduces the supervision cost during the recovery.

9.3. **Future works and Recommendations**

This purposed single degree of freedom exoskeleton device provides guidance and support to a specific, task-based trajectory only. Thus, the patient cannot practice different clinical trajectory required for different disorder or injuries, like extension and flexion of the knee joint and other combinations of these motions. To make this device compatible for different range of motion, design modification is necessary, such as adding one more degree of freedom to produce the different paths and to increase the workspace to accommodate more patients with different level of disability.
REFERENCES
REFERENCES


APPENDIX
Appendix

1. Matlab model for the recovery tracking
"clear all
clc
B= 81000 ;% total body weight in gram
W= 0.0618*B ;% weight of of the lower leg and foot combined
Vout=0;
Vmax=0;
Fmax=0;
Rmax=0;
a=arduino;
    i=0;
    for i=1:1:200;
        Vout(i) = readVoltage(a, 'A0');
        F(i)= 4.970873 * ((Vout(i))^(4.312619));% force applied by exoskeleton to the
        R(i)= ((W-F(i))/W) *100;
        plot(Vout)
        hold on
        plot (R)
        if (F(i)>Fmax)
            Fmax=F(i);
        end
        if (R(i)<Rmax)
            Rmax=R(i);
        end
        if (Vout(i)>Vmax)
            Vmax=Vout(i);
        end
    end
 fprintf('Maximum Voltage output = %d
',Vmax);
% F= 4.970873 * ((Vmax)^(4.312619)); % force applied by exoskeleton to the
fprintf('Maximum force applied by exoskeleton to the leg = %d
',Fmax);
% R= ((W-F)/W) *100; % Recovery in percentage
% if (R>100)
%    R=100;
% fprintf('congractulations you are fully healed')
% end
fprintf('Recovery percentage =%d
',Rmax);