Design of a Low-Cost Lower Limb Rehabilitation Exoskeleton System

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Abstract. Human locomotion is probably the single most crucial thing to carry out daily activities in a person's life. It is hence a necessity than an option to help people walk, when they lose that essential ability, for any reason. Rehabilitation is a methodical process, usually carried out by trained professionals in this regard. With the current lifestyle, the increase in factors leading to disabilities is on the rise, and there is a pressing need to rehabilitate the needy. With a lack of specialized medical facilities and personnel to meet the rising demands, especially in many rural parts of India, a low cost rehabilitation device is necessary. The devices that exist for this purpose remain inaccessible to the masses mainly due to their cost, availability, and ability to adapt to different patients. In many of the works from literature, it is seen that actuators play a significant role in driving the costs high. Also, the designs need considerable customization for every individual patient, which makes mass production difficult. In this work, an affordable lower limb exoskeleton system is designed. This device is adaptable to the varying anthropometry of the Indian population while being affordable by the masses. Electric linear actuators are used to keep the costs low, with a simple control system. Adjustability and modularity are maintained throughout the design as an essential criterion.

1. Introduction
Walking disabilities constitute a significant concern in the world today. In a developing country like India, it is seen that the movement disability accounts for 20.2% of all disabilities, according to the 2011 census. House to house surveys put this number at 31.4% [1]. Not being able to walk poses many problems to human beings, firstly, inability to perform day to day chores, being unable to move. Secondly, being confined to bed causes issues such as osteoporosis, pneumonia, etc. Lastly, it can lead to psychological trauma, which can drive the patients’ and their caretakers’ morale down [2]. Hospital-based rehabilitation in physiotherapy has its high perks due to constraints such as costs, equipment availability, trained personnel, etc. [3].

A significant contributor to movement disability is stroke [4]. WHO has published a set of guidelines called the STEPS Guidelines, followed in a good number of surveys in India and abroad. These surveys observe that India is in the middle of a stroke epidemic [5][6]. The rehabilitation services in India are not up to the needs of the country [2]. The risk factors that lead to stroke are on the rise [7]. After Covid – 19, doctors have observed a rise in stroke cases [8] [9]. However, this needs to be ascertained by scientific studies. Walking disabilities are also caused by Osteoarthritis (OA) and Cerebral Palsy (CP) [10] [11]. Aging causes primary OA, which is the most commonly seen form of OA. Other diseases or conditions can cause secondary OA. The male population has a higher risk of suffering from OA (males – 31.6% and females – 28.1%). A physically demanding work style such as that of daily wage workers,
significantly lowers the chance of being affected by OA (22.2%) [11]. Abnormalities in brain development, often before birth, lead to Cerebral Palsy (CP). Perinatal asphyxia (brain damage at the time of childbirth) is seen more commonly in India compared to other countries [12]. In 15-20% of children suffering from CP, physical therapy and orthotics can substantially improve their quality of life [10].

An exoskeleton is an external support suit that takes the wearer's load and may be powered or passive. A rehabilitation exoskeleton is a device that can address some of the concerns discussed above, if not all of them. Research on the development of exoskeletons has been going on for decades, but affordability and availability to the masses is still a distant dream in India. The main contributor to the cost would be the actuator used in the devices. In the present work, a low cost, lower limb rehabilitation active exoskeleton, shown in Figure 1, is designed using electric linear actuators. A discussion on the same is presented in this paper.

### 1.1. Human lower limb and gait

The human lower limb consists of four parts, namely the hipbones, thigh, shank, and foot, connected through the hip, knee, and ankle joints, as shown in Figure 2. Each joint is unique and allows different types of motions, actuated through the muscles. These joint motions are studied along biological planes, which hypothetically divide a human body into different segments. The planes are named sagittal, coronal, and transverse planes. The movement of limbs about different joints and along different planes are tabulated in Table 1.

![Figure 1. Exoskeleton system assembled onto a human being. The figure shows each module assembled and strapped onto the human body.](image1)

![Figure 2. Parts of the lower limb of a human being, showing the limbs (thighs, shank and foot) and joints (hip, knee, and ankle)](image2)

| Movement | Movement | Angle from ref (°) | Range of Motion (°) |
|----------|----------|-------------------|---------------------|
| Hip      | Flexion  | 130               | 140^b               |
|          | Extension| -10               |                     |
|          | Abduction| -40               | 70                  |
|          | Adduction| 30                |                     |
|          | Later (external) rotation| -40 | 70 |
|          | Medial (internal) rotation| 30 |                     |
| Knee     | Flexion  | -140              | 140^b               |
|          | Extension| 0                 |                     |
The walking cycle of a human being is called the 'Human Gait Cycle'. Conventionally, it starts with the human being's foot contacting the ground, to the subsequent contact with the ground during locomotion. Researchers use angular and torque sensors to measure the joint angles and torque at each joint during gait, and the data hence published is called the Clinical Gait Analysis (CGA) data \cite{13} \cite{14}. This data is used in the design of exoskeleton systems to know the range of motion and force/torque outputs required from each actuator. The joint angles and torque are measured w.r.t % gait cycle. A sample plot of CGA data is shown in Figure 3\cite{13}:

In most lower-limb exoskeleton designs, it is seen that the gait motion along the sagittal plane alone is considered, as most of the human gait motion occurs along with it. This approach simplifies the design, requiring lesser number of actuators, and hence costs much lesser. This same approach is considered here. However, it is noted that many designs do not allow limb motion out of the sagittal plane. This issue is of significant concern. Human gait varies significantly between each individual and may have out of plane motions, arresting which can cause serious injuries.

An essential requirement in the exoskeleton design is to understand human anthropometry. The limb weights and limb lengths are essential in deciding the length of linkages and for selecting actuators. The limb lengths and weights are represented as percentages of the individual's height and body weight, respectively. The approximate compilation of data on human limb weights and lengths is provided by Aydin Tözener’s work \cite{15}. The approximate range of joint angles are obtained from the CGA data.

2. Research on rehabilitation exoskeleton systems
Research on novel exoskeleton designs has been a hot topic, whether for rehabilitation or for supporting workers' heavy loads. It is seen from literature that in the design of exoskeleton systems, the design of the structure is part of the whole design procedure. The other significant aspect is to design a control...
system, which is a challenge in itself. In the present work, more prominence is given to the design of the structure with a simple control system that can be upgraded later.

Chen et al. [16] designed a device, which is a modular knee-ankle-foot portable anthropometric rehabilitation exoskeleton. Series Elastic Actuators (SEA) were designed to provide mean as well as peak torques. The exoskeleton modules were designed to be used separately or together. It is seen that the SEA used increased cost as well as complexity of the device, with the motor used in SEA being considerably expensive. Wu et al. [17] developed a 3-DOF Lower Limb Rehabilitation (LLR) robot for patients' motion recovery, using a brushed DC motor with an encoder as the actuator. An adaptive sub-controller was designed to mitigate the errors induced by friction and disturbances from the patient. Emergency stop switches were provided for safety of patient. Though the link lengths were kept adjustable, the system used, utilizing screws, was discontinuous, causing misalignments between the device’s joints and that of the human, resulting in discomfort. Accoto et al. [18] designed an exoskeleton called LENAR, which is non-anthropometric. Use of the non-anthropometric arrangement provides not only a large number of linkage options to select and optimize, but also helps to mitigate the joint misalignment issue. It ensures less don and doff times, and the adjustable link fits different users’ anthropometry. Maxon brushless DC motors were the actuators used, with required encoders and controllers. The difficulty in this work arises due to the synthesis of non-anthropometric linkages, which are not as straightforward as their anthropometric counterparts. Hence, it is a necessity more than anything to develop an economical alternate powered exoskeleton for the needy, as physiotherapy and rehabilitation costs for post-medical treatment can push families into economic crisis [2].

3. Materials and methods
In the present work, the intention is to design an exoskeleton system that fits the requirements of the Indian population. There should be a trade-off at places to keep the costs low, but that should not hamper the functionality in any way. This approach poses inherent issues such as the gait data being unique to each individual, joint misalignments causing discomfort, inherent limitations of the modular and unified design approach, user safety, system operation without expert intervention, etc. It is hence required to consider certain requirements in the end product. The device should be a low cost, portable system, which fits different patients, with different anthropometry. The parts designed should be modular and interchangeable which will be used in uniped or biped configurations. The product should be bought off the shelf with no customisation to the individual, and should have minimum maintenance costs, if any. Don and doff time should be kept low. The passive joints are kept passive (free to move) to avoid any injury or discomfort to the wearer. Crutches will be provided with the device to avoid falls and for the patient's psychological benefit. The device will also need relevant sensors and control system, with an emergency stop button provided to the patient or the caretaker, in case of any discomfort.

The first step in designing the exoskeleton system is to select and evaluate the kinematic linkages that is to be used in the present work. The actuators that will be used was initially selected as pneumatic actuators, but due to issues discussed later, electric linear actuators is selected to be used. Interchangeable, modular links were designed and assembled to form the exoskeleton assembly in a CAD software. The model is tested for the necessary load condition and kinematical operation. A simple control system with necessary components is also designed. These topics will be discussed in detail under further headings.

3.1. Linkages of the mechanism
The linkages that is to be used in the mechanism is first selected. The listed linkages are modelled mathematically, to map the desired output, i.e. force and stroke length of actuator used to the known input. These models are evaluated mathematically, and the optimal linkage for each joint is selected.

3.1.1. Kinematics and kinetics of mechanism- Synthesis. The linkages that attach to the human limbs are either anthropometric or non- anthropometric; in a sense, it can mimic the human limbs' lengths and joints, or it may not. The latter design is challenging to synthesize, as was in the case of LENAR
[18]. However, it avoids discomfort to patients in case of misalignments. If a rotary actuator is to be used, the mechanism is straightforward, as the actuator is placed directly at the joint and provides a rotary motion, but these motors are expensive. The difficulty arises when the actuator used is a linear one. Mechanisms to convert the linear motion of the actuator to rotary motion at the joint, are plenty. The most optimal linkage may be selected based on the joint's specific requirements such as space constraints, torque requirements, etc. In this regard, five linkages (including one with a rotary actuator) are shown in Figure 4, along with their simplified diagrams, created using SolidWorks software.

![Kinematic linkages](image)

**Figure 4.** The kinematic linkages, along with their simplified versions of linkages for linkages (a), (b), (c), (d), (e). Linkage (a) uses a rotary actuator placed in line with the biological joint. Drawings were created utilizing SolidWorks software.

The important criterion in selecting a linear actuator, be it pneumatic or electric, is to specify its force and stroke length. However, the torque and angle at the joint are the known quantities. By using principles of trigonometry, equations are derived for the force and displacement of the actuator, knowing the torque requirement and angular displacement at the joint. The stroke length is obtained indirectly. For the extreme values of joint angle $\theta$, the values of $a$, which is the distance from the hinge point of the rear end of the linear actuator to the front end mounting point of its slider, is calculated, and the difference of which gives the stroke length ($s$) required. This data helps to select a linear actuator of required specifications. The formulae of $a$, $F$, and other vital parameters are tabulated in Table 2.

**Table 2.** Lower limb joints and their motion.

| Kinematic linkage | $a^b$ | $a \rightarrow \theta$ | $F \rightarrow \tau$ | Other equations |
|-------------------|-------|------------------------|----------------------|-----------------|
|                   |       | $a \rightarrow \theta$ | $F \rightarrow \tau$ | -NA-            |
|                   |       | $\frac{r}{h \sin(a)}$  | $\cos(a) = \frac{a^2 + (h - \frac{e}{\sin(\theta)})^2 - (b - \frac{e}{\tan(\theta)})^2}{2a(h - \frac{e}{\sin(\theta)})}$ |
\[
\begin{align*}
\text{c} & \quad \sqrt{(h \cdot \sin(\theta - \gamma) + b)^2 + (h \cdot \cos(\theta - \gamma) - e)^2} \\
\text{d} & \quad \sqrt{(b - (h \cdot \sin(\theta - \gamma))^2 + (e - (h \cdot \cos(\theta - \gamma))^2} \\
\text{e} & \quad \sqrt{(-p \cdot \sin(\delta' + \theta) + b)^2 + (-p \cdot \cos(\delta' + \theta) - e)^2}
\end{align*}
\]

\[
\begin{align*}
\tau & = \frac{\cos(a)}{h \sin(a + \gamma)} \\
\alpha & = \frac{(\theta - \gamma) + \sin^{-1}\left(\frac{(b - (h \cdot \sin(\theta - \gamma))}{a}\right)}{2ad} \\
\alpha' & = \frac{\tan^{-1}\left(\frac{d}{c}\right)}{\sin(b)} \\
\cos\beta & = \frac{b^2 + e^2 - a^2 - c^2}{2 \cdot a \cdot p}
\end{align*}
\]

\(^a\) Stroke length, \(s = a_1 - a_2\), where \(a_x\) = value of \(a\) at \(\theta_x\)

\(^b\) For linkage (a), since a rotary actuator is used, directly, the required \(\theta\) and \(\tau\) are obtained from the input data.

3.1.2. MATLAB Results. The optimal kinematic linkage for a joint is selected using the equations discussed in Table 2, evaluated for different input conditions using MATLAB software. Here, the parameters such as the length of the limbs and other geometric dimensions are kept constant, based on human anthropometry and other physical constraints. The value of joint angle (\(\theta\)) and torque (\(\tau\)) are also necessary for computations. Two approaches are taken in this regard. The first is to directly use CGA data, which gives the said variables as a function of the \% gait cycle. The other is to vary the joint angle based on the range of angular motion desired at each joint and evaluate the torque using a simple relation. If the limb has a mass of \(m\), length \(L\) and for a given joint angle \(\theta\),

\[
\tau = m \cdot g \cdot L \cdot \cos\theta \quad (1)
\]

Using both the methods, each type of linkage was evaluated for the three joints, and the results obtained included plots of \(F\) and \(a\) w.r.t \% gait cycle, as well as the maximum values of force and stroke lengths. However, linkage (e) was tested only for the hip joint, as it was considered out of necessity when it was found that the other linkages would not fit at the hip joint due to space constraints. The CGA data were utilized from a data set prepared by the research of Bovi et al. [14], who has compiled a data set from a vast number of trials on healthy adults and has provided the mean and standard deviation values of the parameters against percentage gait cycle. The forces and stroke lengths' peak values obtained are tabulated in Table 3, and the rows which are in bold are the linkages that were selected to be used in the final design. A very similar process was also carried out to find the velocity and acceleration required out of the actuator's slider when the joint angular velocity and angular acceleration were provided, by drawing generalized velocity triangles.

**Table 3.** Summary of results of \(F\) and \(a\) for different joints and linkages

| Linkage | Force (N)  | Stroke (mm) |
|---------|------------|-------------|
|         | CGA based  | Limb weight-based | CGA based  | Limb weight-based |
| b       | Hip        | 261.2458    | 81585      | 35.7456      | 38.2510      |
|         | Knee       | 485.3332    | 674.7398   | 105.0907     | 108.7687     |
|         | Ankle      | 3532.2      | 558.8533   | 46.5096      | 47.8340      |
| c       | Hip        | 199.4437    | 408.5012   | 107.6324     | 109.9089     |
|         | Knee       | **224.6899**| **139.9799**| **140.3408**| **131.6203**|
|         | Ankle      | 744.6044    | 111.4417   | 64.3778      | 70.8693      |
3.2. **Exoskeleton actuators.**

Actuators play a vital role in the exoskeleton system, as they provide the force required for locomotion. It was noted from literature that selection of an economical actuator is vital in keeping the costs low. In this regard, two types of actuators are discussed below.

3.2.1. **Pneumatic cylinders.** For reasons mentioned above, it is preferred to select a linear actuator and not a rotary actuator, even if it means the added challenge of controlling the angular rotation resulting from the linear motion from the actuator. The two linear actuators that were of interest were the pneumatic cylinders and the electric linear actuators. Hydraulic system, though better in terms of force output obtained, presents challenges such as cost, complexity, and weight. Pneumatic cylinder was the actuator of choice initially in the design procedure. To select a pneumatic cylinder, the parameters to be selected are the stroke and the bore. Stroke required is directly taken for each linkage arrangement, referring to Table 3 and bore can be calculated or referred from the catalog (of a commercially available pneumatic cylinder) for the required force and for a suitable pressure input. The pneumatic cylinder hence considered was a 25mm bore and 160mm stroke pneumatic cylinder. By knowing the speed required from the pneumatic cylinder, the required flow rate was calculated. For multiple cylinders connected to an air supply tank, the capacity required for a reasonable operation period can then be determined.

As shown in Figure 5, the primary pneumatic circuit consists of an air tank (reservoir), which connects to a 4/3 DCV (directional control valve). The other port of the DCV connects to an exhaust. The outputs of the DCV connects to the two ports of the pneumatic cylinder. Throttle valves are used to control the speed of actuation. The DCV is operated by a solenoid, which in turn is controlled through Arduino. Some inherent problems to note here are the high speed of actuation, which is very difficult to precisely control, higher pressure required for actuation, and noisy operation.

![Figure 5](image_url)

**Figure 5.** (a) The pneumatic circuit for one leg, and (b) symbols used in the circuit. The circuit consists of three pneumatic cylinders, each operating a limb on one leg, either the thigh, shank, or the foot. A similar circuit will be used for the other leg, with three more cylinders. Illustration created using Festo Fluidsim software.

It was calculated that the cylinder's capacity required to operate the exoskeleton for one hour is around 100 liters, which is quite impractical, if not impossible. It is possible to use a mid-sized
compressor to keep the tank filled and operate the exoskeleton continuously, but those compressors are not portable. A low flow rate, small compressor can fill the reservoir of 25ltr in about 10 min, but keeping it filled at high pressure is not possible as these compressors run only for about 10 min at a time. These issues and impracticalities, along with the ones discussed earlier, makes pneumatic actuators a less feasible option.

### 3.2.2. Electric linear actuators.

Electrical linear actuators are used in many industrial applications. They essentially are identical to the pneumatic cylinders when it comes to installation in the exoskeleton system and hence requires a minimum redesign of the device structurally, if any. They are directly powered from a 12V/24V DC power supply and can operate for long durations. These actuators comprise of a DC motor powering a lead screw with a gear train in between. A linear actuator providing around 500N force and a stroke of 150mm is selected for the present application. An issue with the linear actuators in general, is the speeds of actuation, as they are usually slow (nearly 7mm/s). However, models with faster actuation speeds are available at the cost of peak force while still meeting the force requirements of this design. The cost is higher than that of pneumatic cylinders but it is justified since the costs of reservoir, compressor, valves, and tubing are not involved. Hence, an electric linear actuator fits the requirements of the present design satisfactorily. A pneumatic cylinder and a linear actuator are shown in Figure 6. The linear actuator shown here is a Visesh Transmissions - LAG 01, with a load capacity of 50 kg (500N) and a stroke length of 150mm.

![Figure 6. (a) Pneumatic cylinder and (b) Electric linear actuator, LAG 01 from Visesh Transmissions.](image)

#### 3.3. Model of the rehabilitation exoskeleton

Bearing in mind all the discussions so far, the exoskeleton model is virtually built using a CAD software, Unigraphics NX 10. Aluminum fencing section (30*30, Al 6063) and aluminum box tube (1.5", square, 3mm wall) will be used to form a length-adjustable link. Aluminum fencing sections provide the advantage of easy and rapid fabrication and modifications. Mild steel and aluminum sheet metal will also be used. Modularity and interchangeability are maintained throughout the model. For the joints between links, rod end bearings (used in automobiles, like go-karts) are used to allow out of plane rotations.

#### 3.3.1. Limb modules of the exoskeleton

A linkage that fits parallel to the human limb, with all the parts and actuators required to operate it, is called a limb module. It is so designed as to have minimal differences between each module and to keep the modules interchangeable. Each module of the exoskeleton essentially consists of a length-adjustable linkage, with mounting points for the rod end bearing. An L-shaped extension is attached to the linkage, in different configurations, based on which limb it is used on. There would be a provision to attach the rear end of the actuator onto the linkage. The actuator's moving end is attached to the L-shaped extension, with rotation allowed at both ends of the actuator. The limb module is shown in Figure 7(a). Aluminum is chosen as the material for all parts due to its lightweight and ease of fabrication where fencing sections are used. The module was subjected to structural analysis using Ansys structural, based on the load conditions expected on it, and was found to be safe. The Von-Mises stress on the module is shown in Figure 7(b).
Various concepts have been considered for the design of the limb module. Different parameters were changed between each concept to select the optimal module configuration, the parameters being the type of actuator, hinge, and method of adjusting the link length. A total of ten such concepts were considered. They were systematically evaluated, and a concept that uses a linear actuator, connected with a simple hinge, and having a length-adjustable link, made with an aluminum fencing section sliding in an aluminum box tube has been selected as the most optimal design, as shown in Figure 7(a).

3.3.2. Exoskeleton assembly. In the assembly of the exoskeleton system, as mentioned earlier, kinematic linkage (e) is used at the hip joint, linkage (c) at the knee joint, and (d) at the ankle joint, demanding different arrangements of the L-shaped extension. An adjustable back support system is designed to support the patient's as well as battery's load and provide attachment points to the limb modules at hip joint. The back support system and the limb modules are attached securely at respective locations using belts with adjustable tension. A bag pack is provided, which houses the battery and the control circuitry. The assembled exoskeleton system is shown in Figure 8.

Figure 7. (a) Limb module of exoskeleton and (b) Analysis of limb module of exoskeleton under expected load conditions, shows no failure.

Figure 8. Exoskeleton system assembly. The limb modules, in the configurations discussed, are assembled into two sets, one for each leg, and are assembled onto the back support.

To validate if the mechanism can perform the required kinematical task of walking, Unigraphics NX 10 also provides a 'motion' workspace, where such a kinematic analysis is carried out, with the prior knowledge of CGA data and the corresponding value of $a$ of linear actuators. The system provided a reasonable walking motion, as shown in Figure 9. The product is to be fabricated using the above mentioned materials. It was aimed throughout the design process to keep the fabrication process simple.
3.4. Control system and power supply

3.4.1. DH modelling. One of the crucial requirements of a sound control system is to accurately position the end effector, in this case, the foot, at any point in its workspace. This positioning is accomplished by Denavit–Hartenberg (DH) modelling. The simplified model shown in Figure 10(a) is used to obtain a DH model, and a parameter table as shown in Figure 10(b) and 10(c) respectively, and the transformation matrix $T$ is obtained. This result is used to obtain the position matrix of point P in terms of the joint angles and using formulae from Table 2 in terms of actuator extension as well.

\[
P = \begin{bmatrix}
P_x \\
P_y \\
P_z \\
0
\end{bmatrix} = \begin{bmatrix}
L_3 \cos(\theta_1 + \theta_2 + \theta_3) + L_2 \cos(\theta_1 + \theta_2) + L_1 \cos \theta_1 \\
L_3 \sin(\theta_1 + \theta_2 + \theta_3) + L_2 \sin(\theta_1 + \theta_2) + L_1 \sin \theta_1 \\
0
\end{bmatrix}
\]

(2)

Figure 10. (a) A simplified model of a human leg, (b) Denavit–Hartenberg (DH) model, and (c) DH parameter table. Drawings created in SolidWorks software.

3.4.2. Power supply. For an application where the range of walking is minimal, such as while walking on a treadmill or testing the prototype, a Switch-Mode Power Supply (SMPS) of the required specifications may be used. According to the linear-actuator specifications, for a 24V power supply, current $I = 3A$. Since six actuators are needed to operate all the joints, and a current divider circuit is used, $I = 6*3 = 18A$. A 24V, 18Ah battery is used, and an 80% efficient circuit is assumed. Time of operation of the actuator powered by the battery of capacity Ah ampere-hours, drawing current I, can be shown to be:

\[
T = \frac{0.8 \times Ah}{I}
\]

(3)

Substituting the known values, $T$ comes out to be 48 minutes, which is a good duration for the device. The battery weighs around 10kg (5kg x 2), supported by the back support as discussed. Four 9Ah sealed lead-acid batteries may also be used, weighing a couple of kilograms less in total.

3.4.3. Control system. A simple, effective control system is to be designed for the device. The system should independently operate each actuator and verify if the desired angle is reached at the joint, which is set by the biological limits of that joint. The electric linear actuator is essentially a motor powering a lead screw through a gearbox. The speed can hence be varied by using Pulse Width Modulation (PWM). A potentiometer or an absolute rotary actuator is used at the joint to verify if the required angle is
reached. The motor power is cut off / reversed at this point. An Arduino Uno board is used to carry out
the control, interfacing with the motor through a motor controller such as BTS7960 or through a simple
relay board. However, the relay board cannot provide PWM. Load cells placed at the actuator front end
measures the force output from the actuator. An emergency stop button is provided to the patient, which
retracts the device to the neutral position and cuts off the power supply in the event of any injury or
discomfort to the patient. Electromyography (EMG) sensors provide muscle activity information to
the control system. A simple flowchart showing the operation of the control system is shown in Figure
11. A simple plywood mock-up to understand the system's operations, using Arduino, potentiometer,
and a relay board was built and tested for different angular ranges. The mock-up is as shown in Figure
12.

4. Results and discussions
A rehabilitation exoskeleton system that is economical in comparison to existing products is designed,
considering the needs of the Indian population. The product is modular, adjustable, with interchangeable
parts, and uses electrical linear actuators. Further, it is intended to fabricate this design and test it in lab
conditions and on human subjects. EMG sensors and PWM needs to be integrated into the control
system. The modular and adjustable design poses few inherent limitations. It may not be possible to
comfortably fit the device to each individual due to gait and anthropometry variations, which can cause
group misalignments. The motion of limb out of sagittal plane is allowed by the use of rod end bearings,
but there is a limitation to that even then. Usage of linear actuators to produce an angular motion at the
joint is not the most efficient way to produce and control such a motion, and can cause difficulties.

The design's nature provides opportunities for modifications in the future, such as changing the
actuators, designing a much more robust control system, etc. One area to explore is to use pneumatic
cylinders with compressed carbon-dioxide, where large quantities of the gas can be stored in a small
container and hence it can operate the system for longer durations. The limitations have to be mitigated,
and design modifications are to be explored in the future. Through this work, it is intended to make the
exoskeleton technology accessible to those who need it the most.

5. Conclusion
An economic lower limb rehabilitation exoskeleton system was designed, considering the varying needs
of the Indian population. Concepts of modularity and interchangeability was maintained throughout the
design. Different possible kinematic linkages were identified, and their force and stroke requirements
for given angular motion and torque at the joint were mathematically modeled and evaluated using
MATLAB software. Based on the understanding of the kinematics of different linkages, and the space constraints at each joint, an exoskeleton system model was built in Unigraphics NX 10. It was found that the electrical linear actuators were better than their pneumatic counterparts. The requirements of the battery, DH modeling, and a simple control system were also briefly discussed.

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