Mechanical Design of Knee and Ankle Exoskeleton to Help Patients with Lower Limb Disabilities

Ignatius Deo Putranto[1], Eka Budiarto[1], Kidarsa, Lydia Anggraeni[2]
[1]*Swiss German University, Alam Sutera, Indonesia*
[2]*Bioteknik, Bandung, Indonesia*
ideoputranto4@gmail.com, eka.budiarto@gmail.com, lydia@kidarsa.com

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Abstract: In the medical world, exoskeleton in this project refers to an orthosis, which is applied externally to the user's body. This project aims to develop an exoskeleton with an affordable cost especially for assistance and rehabilitation. This project mainly cover around the mechanical design of the product, covering of two primary joints of the lower body part: knee and ankle. The overall size of the exoskeleton referring to limb segments is limited to a select range of subject proportion. Each joint is to be motorized and equipped by certain mechanism such as the application of linear actuator and four-bar linkage mechanism for the knee and ankle joint, that enhances the efficiency with respect to speed and power transmission during actuations, as well as the ability to deliver a smooth human locomotion. The types of body movement discussed is around the sagittal plane which are flexion and extension for knee joint, including the dorsiflexion and plantar flexion of the ankle joint. In order to help the patient regain the ability to move as mentioned, the exoskeleton is made especially for external use with existing limbs for the lower body part to move, as well as to endure the subject's weight.

1 INTRODUCTION

The development of exoskeleton is one of the most progressive topics in this decade, with each step that aims to produce the most accurate lifelike motion that makes the user feel as if it is part of their body (Chen et al, 2015). Essentially, a human walk or moves within the sagittal and frontal plane. The application of exoskeleton is primarily utilized on limbs which handles the person's mobility and stability. The means of an external actuations that produce a movement on specific limbs, such as the leg and arm, enhances the power of the corresponding joints. Some of the applications are widely used in the medicinal branch of orthotics that aids people in moving their limbs as a form of rehabilitation during their recovery. To be specific, this project will prioritize on the support for the knee and ankle joint.

In this current project, the main actuator to simulate the knee and ankle joint will be using a motor-powered linear actuator mechanism. To achieve a more flexible and convenient design the joint mechanism should mainly be considered, with the fact that an actual knee joint has a slight displacement that affects the shank (Wang et al, 2011). The hypothesis of this project is that the mechanical design which includes linear actuator and four-bar linkage could produce an efficient in terms of strength and speed motion for the exoskeleton to mimic human gait in terms of joint angular motion.

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Table 1: Subject's Inclusion and Exclusion Criteria.

Inclusion	Exclusion
• This exoskeleton was aimed towards more flaccid, hypotonic lower extremity function. • The subject is unable to stand and walk on his own due to weakness or partial paralysis of leg due to Spinal Cord Injury, Spinal Muscular Atrophy, Spina Bifida, or Polio. • Good trunk control. • Preferably, the subject should be able to stand on their knees and take a couple of steps on their knees.	• Pain in ankle and/or knee joint when articulated. • Fragile skin integrity and skin vulnerability. • Poor trunk control. • Moderate to excessive spasticity. • Complete paralysis of lower extremity. • Fragile bone such as in the case of osteogenesis imperfecta.

2 METHODS

The design overview of the knee and ankle is composed of a joint that connects thigh and shank body part. The four-bar linkage mechanism is applied for the knee joint as it is efficient and simulates the joint motion and muscle contractions in the knee (Widjanarko, 2018).

The scope for this project is to implement a motorized actuator in that case such as a DC Motor with a rotary to linear output conversion by the means of lead screw. The chosen actuator is expected to be able to handle the maximum load subjected to it through calculation and testing. For the knee joint specifically, the actuation will be conducted using four-bar linkage mechanism.

The production process starts with the subject measurement with provided inclusion and exclusion criteria. However, the subject measures are only gathered to specify the user requirement, and is not intended for direct testing. Followed by gait cycle analysis to measure the maximum and minimum angular range of motion on each joint, also to compute the needed torque and speed respectively.

2.1 Subject Measurement

Before proceeding to the next process, subject measurement is crucial to determine the limitation scope that includes load, body segment lengths and the range of adjustments that the final product must satisfy. The subject is not limited by a certain range of age; however, the subject must satisfy the criteria shown in Table 1. The reason for these exclusions is to prevent unexpected injuries or complication on the subject, if the exoskeleton would be tested on a real subject.

The data will be compared between real life measurement and the calculation using the referenced body segment ratio (Winter, 2009). The reference will be useful to estimate the segment center of mass, weight ratio and to compare the length differences of each segment.

Table 2: Body Segment Length.

Body Segment	Length (cm) Real Measurement	Length (cm) Reference
Thigh	28.910	34
Shank to Foot	33.630	28
Shank	29.028	23
Foot	17.936	19
Waist	22.538	26

Referring to Table 2, the subject was measured on April 8, 2019, with the height of 118 cm and weighing 27 kg. The ailment should also be noted, which is Spinal Muscular Atrophy, affecting the lower limbs. The body segment weights are shown in Table 3.

Table 3: Body Segment Weight.

Body Segment	Weight (N)
Thigh	26.487
Shank	12.316
Foot	3.841
Upper Body (HAT)	179.582

2.2 Gait Cycle

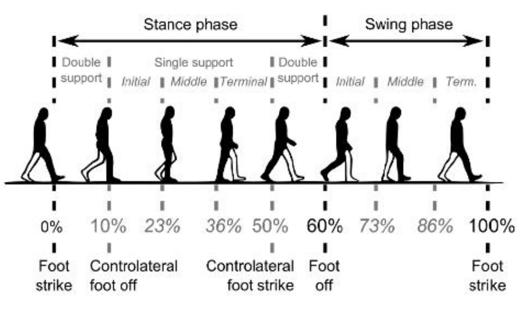

Figure 1: Human Walking Gait (Bonnefoy-Mazure, 2015).

In general, humans walk and run in a common pattern. A single movement sequence as seen in Figure 1 above is called as the human gait cycle which is divided into two phases; the swing phase and the stance phase. These phases are indicated when both feet are on the ground (stance phase) and when one foot is lifted from the ground (swing phase). Considering the swing phase, in which one foot is floating, means that the other limb will act as a support before the person reach their next footing during the swing phase. The cycle continues by the switching between the left and right limb alternatively, performing the walking or running pattern (Kharb et al, 2011).

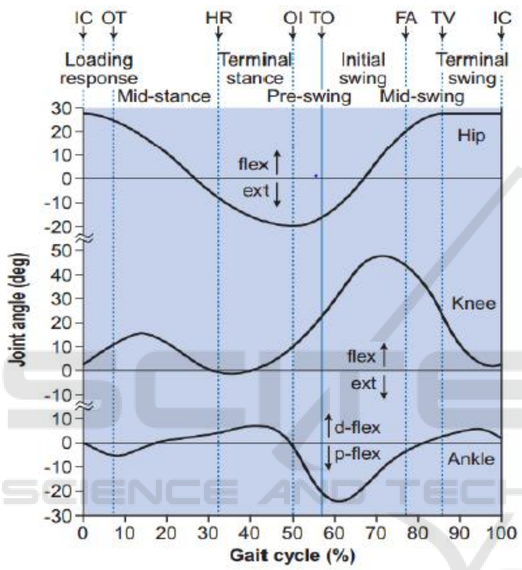

Figure 2: Joint Angle Progression Graph (Whittle, 2007).

The joint angle progression will refer to Figure 2. The graph will be reverse engineered and estimated with 0.5º tolerance. According to this data, the minimal and maximum angle for each knee and ankle joint respectively are 0º until 47.5º and -22º until 8º. The polarity of the ankle angle refers to the dorsiflex (+) and plantar flexion (-).

2.3 Actuation Speed Calculation

The actuation speed also plays a role on the overall machine performance. Each joint speed could be estimated using the data from Figure 2 in a single gait cycle.

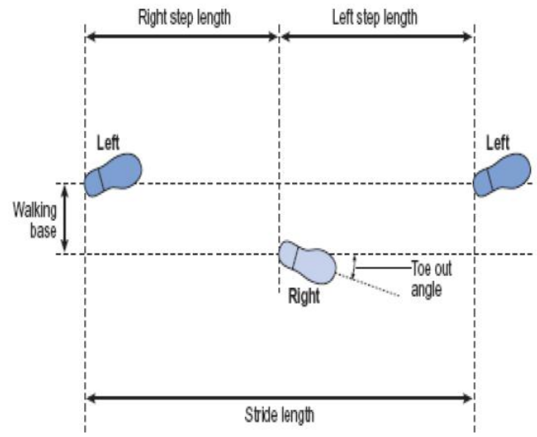

Figure 3: Walking Steps Parameter (Kharb et al, 2011)

The parameter as shown in Figure 3 such as the stride length will be used to calculate the cycle time (s) and walking speed (m/s) of a person. And the calculation for these two variables are as follows:

$$\text{cycle time }(s) = \frac{120}{cadence}\left(\frac{\text{steps}}{\text{min}}\right) \#(1)$$

$$Speed\left(\frac{m}{s}\right) = stride\ length\ (m) \times\ cadence\left(\frac{steps}{min}\right) \#(2)$$

$$Speed\left(\frac{m}{s}\right) = \frac{stride\ length\ (m)}{cycle\ time\ (s)} \#(3)$$

Each person has a distinct stride length on each gait cycle, even with the same walking speed. Therefore, the controlled variable in this calculation are the walking speed = 1.4 m/s (AVG human walking speed) and stride length = 0.8 m. From these values, we could acquire the cycle time from the equation (3), cycle time = 0.571.

Regarding the chosen subject is a child, approximately nine years old, the controlled parameters will be halved, however according to equation (1), the cycle time remains the same. The angular speed parameters are determined from the graph in Figure 2, which states the gait cycle progression (%) and joint angle (º). The range is determined from the steepest slope, which means the fastest speed possible.

2.3.1 Required Time and Angular Speed Equation

$$T_j = (GP_1 - GP_0)\ \times\ cycle\ time\ (s)\ \#(4)$$

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$$\omega_j = \frac{\theta_1 - \theta_0}{t_j} \#(5)$$

From equation (4) and (5), the minimum required velocity of each joint is provided in the Table 4.

Table 4: Joint Velocity.

	KNEE JOINT	ANKLE JOINT
θo (°)	0	5
θ1 (°)	70	-23
Gait 0 (%)	35	45
Gait 1 (%)	70	60
Time in interval (s)	0.19985	0.08565
ω (rad/s)	4.146	5.703

2.4 Material Selection

Stainless steel pipes will be the primary material used for the frame, and PETG for motor mountings. Stainless steel was chosen due to its corrosion resistance and high tensile strength. While the remaining concern is the weight, the material will take form of a pipe to reduce volume. As for the motor mountings, PETG was chosen due to its manufacturing capability through 3D printing that could produce a more complex and flexible design.

The stainless-steel type that will be used as the exoskeleton frame is 304, as it is one of the most common types available in general stores. Even though stainless steel is widely known for its high corrosion resistance, the design and manufacturing process should always be monitored (Vaghani, 2014).

For 3D printed material, the PETG was chosen as it has high ultraviolet resistance in comparison to ABS and PLA. This property would allow the exoskeleton for outdoor use, with minimum probability of deformation.

2.5 Four-Bar Linkage

Four-bar linkage mechanism is efficient, provides a smooth motion and is known to be similar with the actual knee muscle mechanism (Figure 4).

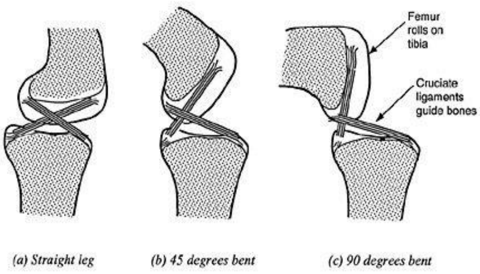

Figure 4: Human Knee Joint.

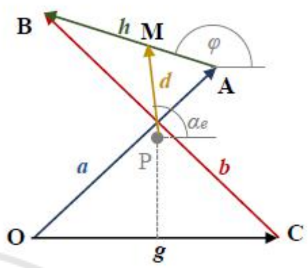

Figure 5: Four-Bar Linkage Properties 1 (Widjanarko, 2018).

Represented as a model in Figure 5, the knee muscles that crosses together are link a and b. In the actual contraction, however, link *a* and *b* changes length because of the flexing. By implementing this mechanism, linear displacement and angular displacement does occur during the locomotion. Such that it simulates the muscle contraction of the knee muscles, these displacements can be estimated and minimalized to obtain an accurate gait performance and ultimately bestows comfort and safety for the user, by adjusting the lengths of link a, b, g and h. These links remain a constant. The link g will be used as a reference point of the system, representing the thigh distal end. As well as link h, representing the shank proximal end.

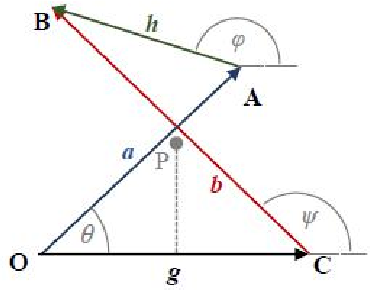

Figure 6: Four-Bar Linkage Properties 2 (Widjanarko, 2018).

The mechanical advantage delivered by this mechanism may also be noted to produce the highest efficiency, which primarily relates to the system angular acceleration. As can be seen in Figure 6, the output angle (φ) can be determined by two other variables which are the input (θ) and follower (ψ) angle using equation (1).

$$\varphi = \arctan\left(\frac{b \cdot \sin \psi - a \cdot \sin \theta}{g + b \cdot \cos \psi - a \cdot \cos \theta}\right) \#(6)$$

The shank angle with respect to the thigh will refer to the output angle produced by this mechanism.

2.6 Mechanism Evaluation

A linear actuator is a term in which the output of a motor is designed such that it is converted into linear motion. Instead of producing a torque, a linear actuator produces force to directly push an object. A linear actuator working principle simply lies to which type of mechanism it is applied (Budynas, 2008), the more common and less complex mechanism is by using a leadscrew. The type of motor used for a linear actuator is not critical as it is only an output converting mechanism. The draw back of a linear actuator would be the material used.

Table 5: Leadscrew Types.

Thread Form	Figure	Uses
Unified		General use.
Metric		General use.
Square		Ideal thread for power transmission.
ACME		Stronger than the square thread.
Buttress		Designed to handle heavy forces in one direction (e.g. truck jack).

There are four types of leadscrew as shown in Table 5. The most common lead screw type available in the market is ACME thread, which has a distinct trapezoid thread. The torque calculation needed to rotate a lead screw mainly refers to these types. For an ACME thread, the torque required to rotate this lead screw is slightly larger than the other counterparts, hence the larger contact area of an ACME thread. The required torque is also based on the orientation of the linear actuator.

2.6.1 Linear Actuator Torque

The amount of torque required to drive a leadscrew to push a certain force is shown in equation (7).

$$T_R = \frac{F \cdot d_m}{2}\left(\frac{l + \pi f d_m \sec \alpha}{\pi d_m - f l \sec \alpha}\right) + T_c \#(7)$$

F = Amount of force to be actuated
d_m = The outer diameter of a leadscrew
l = lead, equals to the number of start times pitch length
f = friction coefficient between leadscrew and material
α = ACME thread trapezoid angle

Equation (7) computes the amount of torque with a direction of force against gravity, in other words the required torque to raise the load. In the other hand, the torque to lower the load is shown in equation (8). A self-locking term could also occur when the required torque is minus.

$$T_R = \frac{F \cdot d_m}{2}\left(\frac{-l + \pi f d_m \sec \alpha}{\pi d_m + f l \sec \alpha}\right) + T_c \#(8)$$

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Commonly, an extra component called as thrust collars are used in a leadscrew mechanism which are attach to the load end of the leadscrew to lift it. Since this exoskeleton's linear actuator design does not apply this function, T_c is zero. The constant values are listed below,

d_m = 8 mm
f = 0.25
l = 8 mm
α = 29º

These constants are related to the dimension, type and material of the leadscrew.

2.7 Frame Model

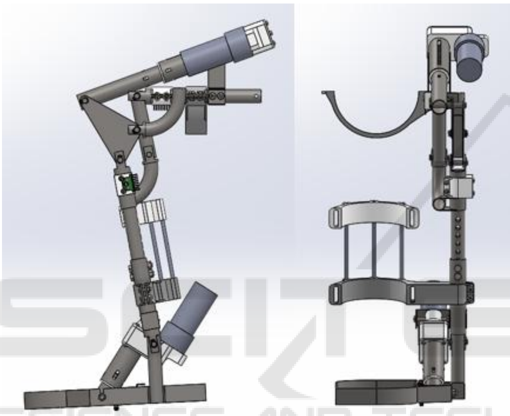

Figure 7: Exoskeleton Solidworks Model.

The frame model is shown in Figure 7. As mentioned before, this exoskeleton is classified as an orthosis which is externally equipped to the subject. The frame would take an approximately 10 cm space of the outer sides of the subject's leg. To attach the exoskeleton, the frame is equipped with a harness fabricated through 3D Printing. For the foot model, the frame was designed suitably for the subject wearing a shoe. The four-bar linkage links was designed to be assembled in two different planes to avoid collision between the parts.

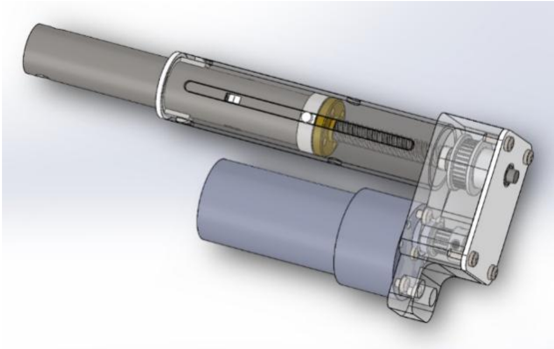

Figure 8: Linear Actuator Model.

The linear actuator design is shown in Figure 8, equipped with a belt and pulley mechanism to increase the output torque ratio. The rotary motion is converted to linear due to the leadscrew rotation causing the nut to drive back and forth. Both linear actuator for knee and ankle joint are positioned for each knee and ankle joint with a pivoting point and custom displacement range. The pivoting point allows an angular motion of the actuator during extension and retraction. The displacement range for each knee and ankle joint respectively are 8.9 cm and 0.75 cm. The four-bar linkage mechanism is only applied for the knee joint instead of both, the reason being that the knee joint requires a wider range of angular displacement whilst considering the required angular velocity shown in Table 4.

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Table 6. Four-Bar Linkage Properties.

Link A	Link B	Link G	Link H	|A-B|	|G-H|	P_x	P_y
(mm)							
41	71	40	35	30	5	20	30

θ_{min}	θ_{max}	ψ_{min}	ψ_{max}	Φ_{min}	Φ_{max}
(°)					
70.78	148.41	149.8	190.66	4.91	-81.53

The link sizes of the four-bar linkage are shown in Table 6. The minimum angle for both input and follower angle can be obtained from Solidworks. To compute both angles with respect to the actuator displacement is shown in equation (9) and (10).

$$\theta = 70.7811° + \left(acos\left(\frac{229.113 - (7.8 + d)^2}{189.826}\right) - 27.5679° \right) \#(9)$$

$$\Psi = \text{acos}\left(\frac{7097 - 3280cos\theta}{142\sqrt{3281 - 3280cos\theta}}\right) + 180 - acos\left(\frac{3200 - 3280cos\theta}{80\sqrt{3281 - 3280cos\theta}}\right) \#(10)$$

The numbers are in accordance to the segment sizes of the frame and solved using trigonometry. As can be seen in equation (9). the follower link is dependent to the input link. Using the equation of output angle (10)., the output angle can be computed using the variables in Table 6.

Represented as Φ, the output angle at maximum is -81.53º deducted from the calculation as it changes into a different quadrant, while the real maximum value of the output angle is to be added by 180º. Therefore, the angular movement range of the knee joint is 93.56º by design.

As for the ankle joint, by using the same trigonometry method, the relation between the ankle angle and the ankle linear actuator displacement is shown in equation (11).

$$\theta = 8.18° - \left(acos\left(\frac{8600{,}859 - (106.814 + d)^2}{4951.128}\right) - 124.556°\right) \#(11)\#$$

The maximum dorsiflex angle is 8.18º and for the plantar flexion is 21.71º (-).

2.8 Torque Calculation

Linear actuator converts rotary motion to linear motion; therefore, the required torque of the motor is based on the amount of force subjected to each linear actuator.

2.8.1 Knee Joint

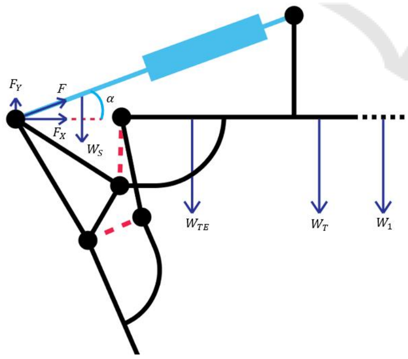

Figure 9: Knee Joint Frame Free Body Diagram.

The knee joint free body diagram is shown in Figure 9. The force that is going to be calculated is the static force during a sitting position which is presumed to have the highest amount of force. The variables required to find F are listed below,

W_s = Weight of LM inner shell

W_{TE} = Weight of Thigh Exoskeleton (Includes one harness, MPU mount)

W_1 = Weight of Upper Body

W_T = Weight of Thigh

The weight of the exoskeleton was estimated using Solidworks, while the weight of body segments refers to Table 3.

$$\downarrow F_y = 0;$$

$$F_y = W_s + W_{TE} + W_1 + W_T \#(12)$$

$$F = \frac{F_y}{\sin\alpha} \#(13)$$

Through static force equilibrium (12) and equation (13) the maximum amount of force acting on the knee linear actuator is 351.454 N. The angle α is 19.95º obtained from Solidworks.

As the acting force has been computed, the torque to raise and lower the load respectively are, 0.934 Nm and -0.042 Nm.

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2.8.2 Ankle Joint

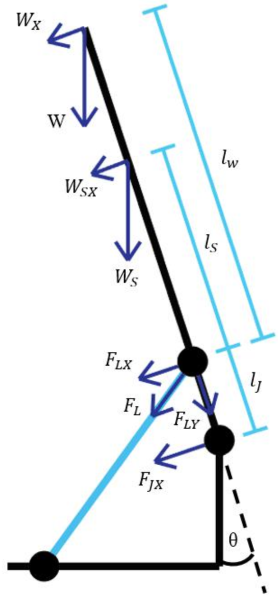

Figure 10: Ankle Joint Frame Free Body Diagram.

The ankle joint free body diagram is shown in Figure 10. The force is calculated during terminal stance when the angle θ is at 8.18º. The variables required to find F are listed below,

W	=	Weight of the upper body, one thigh, and one leg in addition to the weight of the Knee LA, Thigh Exoskeleton
W_s	=	Weight of Shank in addition to shank exoskeleton frame
F_l	=	Amount of force the Ankle LM must endure
F_j	=	Amount of force affecting the ankle joint
l_w	=	W arm length
l_j	=	F_j arm length
l_s	=	W_s arm length

$$\downarrow M_{LM} = 0;$$

$$0 = W_x \times l_w - F_{jx} \times l_j + W_{sx} \times l_s \#(14)$$

$$\uparrow F_x = 0;$$

$$0 = W_x + W_{sx} + F_{LMx} + F_{jx} \#(15)$$

$$F_{LM} = \frac{F_{LMx}}{\sin\alpha} \#(16)$$

Using the moment and force equilibrium method (14) and (15), the force subjected to the ankle linear actuator can be calculated with equation (16) that is 358.505 N with α (the angle between linear actuator and shank) at 55.86º.

Therefore, the required torque to raise and lower the load for ankle linear actuator respectively are 0.953 Nm and -0.043 Nm.

2.9 Inertia Calculation

During swing phase the actuators needs to withstand the inertia of the bottom half of the leg, which is mostly subjected to the knee joint actuator. To calculate the inertia, the equation is based on a simple pendulum inertia,

$$I = m_{sf} l_{sfg}^2 \#(17)$$

The m_l refers to the mass of the both shank and foot (m_{sf}), and r_l is the centre of gravity of the shank and foot (l_{sfg}). The center of gravity can be calculated using equation (18).

$$l_{sfg} = \frac{m_b \times (0.0465 \times l_{sg} + 0.0145 \times l_{fg})}{0.061 \times m_b} \#(18)$$

Substitute equation (17) in terms of total body mass and height, the inertia equation (19) can be obtained.

$$I = m_b h_b^2 (1.51 \times 10^{-3}) \#(19)$$

2.10 Motor Type Selection

The type of motor chosen for the linear actuator is a DC Motor. According to torque calculation in subchapter 2.8, the torque required is not as demanding as opposed to the speed requirement. Considering the weight of the motor would affect significantly to the exoskeleton load, a compact and lightweight motor is preferable and should be taken into account.

However, a motor with these specification, especially with a compact size, is difficult to find in the market and mostly come with a large dimension. Therefore, one aspect should be sacrificed between speed and torque and for this project, the torque is to be prioritized.

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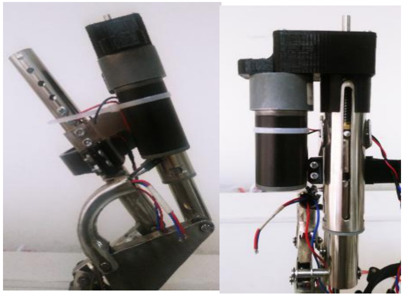

Figure 11: Exoskeleton Final Product.

3 RESULT & DISCUSSION

The final product of the exoskeleton is shown in Figure 11. It weighs 2.055 kg for each leg. The material used for the frame is made of stainless steel. However, the type may vary between 304 and 201 because of manufacturing miscommunication. The pipe shaped stainless steel is also useful for cable management. The foot harness is made from PETG through 3D Printing, the same as the motor mounting. The harnesses are equipped with Velcro to fasten. The shank segment was made to be adjustable, in the range of 5 cm.

Figure 12: Foot Frame.

As seen in Figure 12, the footing is also equipped with Velcro fastener. To the right, the linear actuator of the ankle joint was manufactured as it was designed, connecting the foot segment to the shank segment.

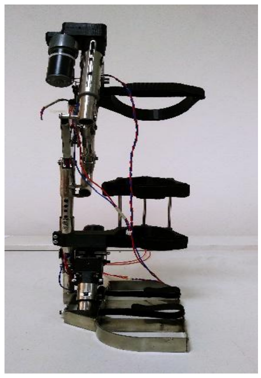

Figure 13: Knee Joint Actuator.

The knee linear actuator connects the thigh segment with the triangle part which drives the four-bar linkage mechanism to rotate the knee joint. Apart from the whole exoskeleton, the knee linear actuator weighs 0.520 kg each, while the ankle linear actuator weighs 0.395 kg each.

3.1 Range of Motion

The range of motion was tested by powering each linear actuator to reach the minimum and maximum linear actuator displacement after product assembly. The angle is the measured using a protractor.

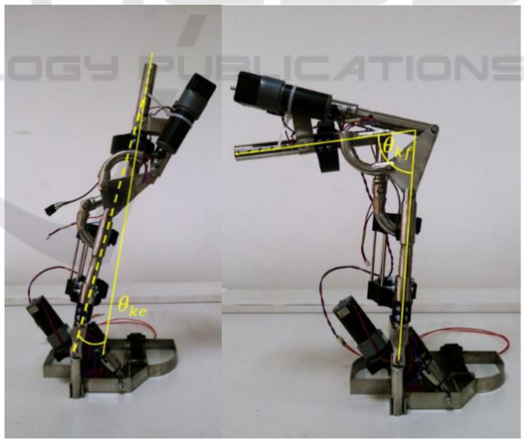

Figure 14: Knee Actuator Movement Range.

Figure 14 shows the maximum extension and flexion angle denoted as θ_{ke} and θ_{kf}. Respectively θ_{ke} and θ_{kf} are 98.5º and 5º, therefore the range of angular motion of the knee joint is 94.5º.

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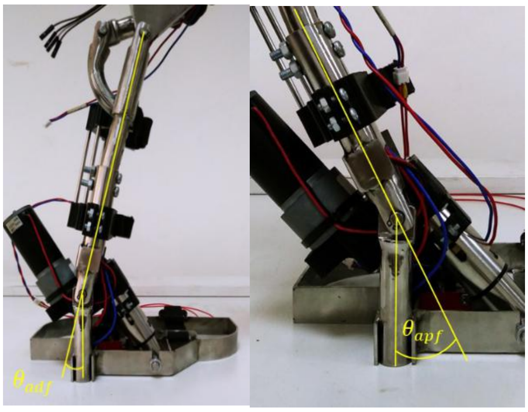

Figure 15: Ankle Actuator Movement Range.

Figure 15 shows the maximum dorsiflexion and plantar flexion angle of the ankle joint, denoted as θ_{adf} and θ_{apf}. Respectively θ_{adf} and θ_{apf} are 15º and 22º, therefore the range of angular motion of the knee joint is 37.

The test result shows an accurate result in comparison to the calculation, especially for knee extension, flexion and plantar flexion of the ankle with a slight error. However, the dorsiflexion angle shows a larger value than expected.

It has been concluded that these errors were the result of manufacturing imperfections in addition to the linear actuator mountings which is made from PETG that could deform far easily. Not to mention that the range of motion was tested without any additional load added to the exoskeleton which may result in a larger error and even reach a breaking point. The adjustment part of the shank segment could also affect the test result due to the pipes having different dimension that creates a gap that causes the frame to tilt approximately up to 3º.

3.2 Load Test

Load testing was conducted to obtain the PWM value when the linear actuator is subjected to a certain load. The PWM value will then be conditioned to the amount of load is being used when conducting the gait analysis. The test resulting the values below is obtained when the leg is in Mid-swing stance. The tests are done using a 24V battery cell.

Table 7: Knee Joint Load Test.

Load	Lowering Torque PWM	Raising Torque PWM	Starting Angle (°)	End Angle (°)
2 kg	34	120	5	87
2.5 kg	40	255	5	80
2.75 kg	30	255	5	61.5
3.25 kg	35	255	5	54
3.5 kg	35	255	5	45
4 kg	35	255	5	44

The PWM value reached its maximum value when 2.5 kg load is subjected to the linear actuator. This is the point in which the end angle is starting to shorten, as more load applied to the device. This value might be affected by the slipping of the timing pulley and the thrusting point shaft not being concentric with the triangle. After certain cycles. The PETG mounting shows a slight deformation, making the inner shell and outer shell tilting.

Table 8: Ankle Joint Load Test.

Load	Lowering Torque PWM	Raising Torque PWM	Starting Angle (°)	End Angle (°)
0.6 kg	58	32	15	35
0.76kg (dummy)	58	36	15	35

The data collected when testing the ankle linear actuator is not as complete as the knee linear actuator. This is due to the slipping of the timing pulley when higher load is added. The 0.6 kg is the weight of the foot frame.

3.3 Actuation Speed Test

The linear actuator speed test was done by incrementing PWM gradually and measure the speed in which the linear actuator reach maximum its maximum range. No additional load was added to the system to confirm the calculation as well. The test was done using a 24V power supply and a motor driver to control the PWM.

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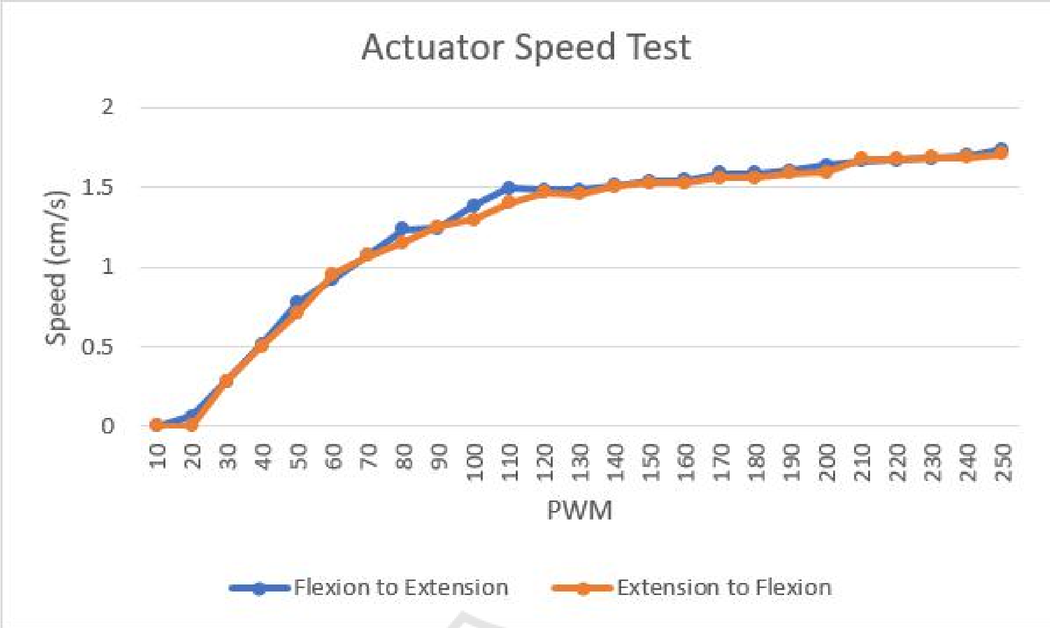

Figure 12: Actuator Speed Progression

Figure 12 shows the progression of actuation speed when given a certain amount of PWM. According to the graph, the speed starts to slowly increase after reaching 140 PWM, meaning the motor is in the range of its rated output power. Notice that there are two curves, this is to show the difference when the linear actuator undergoes a raising torque and lowering torque. The maximum speed that the motor could produce is 1.738 cm/s.

Comparing to the required angular speed of each joint in Table 4, the angular speed produced by the linear actuator can be computed by inverting equation (11). The ankle equation was chosen since the required speed of the ankle joint is bigger than the knee, in addition to the same motor used for all linear actuators. The angular speed produced by the linear actuator is 6.51 times slower than the requirement.

4 CONCLUSIONS

Since the linear actuators are subjected to a large amount of force, different material such as aluminium is recommended to replace the motor mountings. A tensioner would also be recommended if the motor driving has fewer torque than needed running in a belt mechanism.

The data shown in Table 6 nonetheless shows a promising power produced by the linear actuator. However, the speed is much less than the required amount, meaning that the subject could not walk as fast as the human average walking speed. Therefore, this product is more recommended to be used as a rehabilitation or slower activities.

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