Design and Dynamic Modelling of Knee Exoskeleton for Disabled People through ADAMS-Simulink Co-simulation

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ABSTRACT

This paper is written to outline the progress and findings of an undergraduate research project aimed at the designing and modelling of an exoskeleton design of the knee joint for rehabilitation and gait support. It focuses on the rehabilitation potential of the proposed exoskeleton design on patients with Hemiplegia and Monoplegia conditions with the objective of seeking a feasible, simple means of joint actuation to reduce the complexity of the design. Exoskeleton designs are able to provide rehabilitation and improve the overall quality of life of disabled people globally. However, a common issue found almost everywhere is the costly nature of the exoskeletons that are available in the market now as they are exclusive devices, which makes them inaccessible and impractical to the general public, especially in developing countries. Initially, the paper focuses on the review of relevant literature and previous research and evaluating the designs that have been developed as of now. A thorough analysis of the work done by previous researchers and companies was conducted to gather data on the underlying engineering principles and techniques used for the exoskeleton development as well as any limitations or restrictions to the process.

Conceptual designing of the possible solutions was developed using the understanding and knowledge gathered during the literature review. The selection of the best solution was based on the analysis of the pros and cons of all the solutions. The chosen design, utilising a 4-bar mechanism, was then modelled using SolidWorks software to provide a clear visualisation of the system. The kinematic and dynamic analysis of the mechanism was evaluated to analyse the possibility of using the proposed exoskeleton design by replicating the model in ADAMS multibody software. The dynamic analysis was conducted by using the co-simulation platform between ADAMS and Simulink to enable the addition and control of feedback loops within the system. The results of the analysis show that the design can achieve the required motions of the human gait cycle, especially during the swing phase of the gait cycle. The analysis of the actuation torques and reaction forces on the human body showed that an acceptable torque range was possible during the swing phase of the gait cycle.

KEYWORDS: *Exoskeleton, Walking-Aid, Rehabilitation, Multibody dynamics*

1 INTRODUCTION

As evident throughout history, physical disabilities have been a major cause of life-altering situations for people, irrespective of gender or social status. It is estimated that close to 10% of the world population suffers from disabilities. In fact, in developing countries, the prevalence of disability in working-age individuals is estimated to be from 3% to 16% (Mitra et al., 2013).

One main cause of physical disabilities worldwide is paralysis. Paralysis can be defined as the loss of motor or muscle function of a part of the body. In fact, studies have shown that, the number of patients suffering from various forms of paralysis can be over 1.4 million people in Sri Lanka alone (Peiris-John et al., 2014) . The main forms of paralysis are monoplegia and hemiplegia conditions. Monoplegia patients tend to lose muscle function in one limb or extremity of their body while hemiplegia patients usually lose the motor function of a complete side of their body.

Linkages are used in exoskeleton design to transfer the motion from an actuator to the exoskeleton in the desired way. Due to the nature of the linkages, they are promising in being able to reduce the torque required for the exoskeleton actuation and generate motion capable of mimicking the human gait.

As detailed in the research by (Zhang et al. 2019) it is possible to utilize linkage mechanisms, in this case a slider-crank, to provide the actuation torque of the exoskeleton. The actuation is used to generate the required range of motion of the knee joint of the user. It consists of a ball screw connected to a motor that can convert the rotary motion of the motor into a straight line motion. This moves the slider of the slider-crank mechanism which in turn actuates the crank which transfers the translational motion of the slider to the rotational movement required by the user's shank. In this particular design, the slider-crank mechanism is optimized to generate a larger torque at the necessary angles within the desired range of motion of the knee joint. It also discusses how gravity compensation was achieved using springs to reduce the torque requirement of the actuator.

In addition to this, research done by (Gilbert, Zhang, and Yin 2016) provides insight into an exoskeleton that utilizes slider-crank mechanisms at both the hip and knee joint for flexion/extension of the joints while another biped robot design by (Hamon, Aoustin, and Caro 2014) details a novel four-bar mechanism that is capable of allowing a more natural gait motion of the robot. In this case, the 4-bar mechanism is used to mimic the natural 4-bar mechanism found in the human knee joint.

While the motion of the knee joint is controlled through motor actuation, it is possible to drive the motion of the hip joint through passive actuation methods. This passive support could be either to reduce the peak power of the motor by utilising the energy-storing capabilities of springs as detailed in, (Wang, van Dijk, and van der Kooij 2011), by utilizing the springs either in parallel or series to the motor actuator, can allow the peak torque and power values required by the exoskeleton to be decreased. Additionally, spring mechanisms could also be used to counter the weight of certain links in a mechanism, which allows the actuation of that joint with less power. The gravity-balancing of links can be seen in (Arakelian 2016) where a detailed analysis is shown as to how springs can be used for this purpose and also its advantages over other mechanisms such as pulley systems.

Technological advancements and developments have led to the lives of disabled people being improved and the quality of life of these individuals has been retained. However, a common problem faced by many people in low or middle-income countries is the cost of these products. Complexity of the products has also contributed to reduced access to this technology in these countries.

Therefore, it is evident that there is an increasing need for exoskeleton designs that are more suited for these countries by devising designs that are simple, less expensive yet capable of ensuring the necessary assistance required by people with disabilities. The device also needs to use as few actuators as possible in order to reduce complexity and the cost of production.

The loss of muscle activity and strength due to medical conditions can come in two different forms. When the condition leads to weakness of the muscle it is termed as 'paresis' while complete loss of function or paralysis is termed as 'plegia'. When the loss of function is limited to one extremity or limb, it is called 'monoplegia' while the loss of function of one side of the body is called 'hemiplegia'.

This paper focuses on the development of a walking aid design focused on the rehabilitation and gait support of patients with lower-body paralysis, specifically conditions such as monoplegia and hemiplegia. The main objectives of the research are to seek a feasible, simple means of actuation of the exoskeleton, which leads to lower costs and less complexity of the design.

2 METHODOLOGY

2.1 Selection of design and determining linkage parameters

The study generated conceptual designs for the exoskeleton model. The designs were evaluated for their pros and cons to identify the best design for the exoskeleton. Once the design was selected, the viability of the proposed design to be used in an exoskeleton design was validated through the kinematic analysis of the proposed mechanism. The proposed mechanism and the alternative conceptual designs are illustrated in Figure 1 and Figure 2.

Alternative design concepts that were generated are shown in the following figure: -

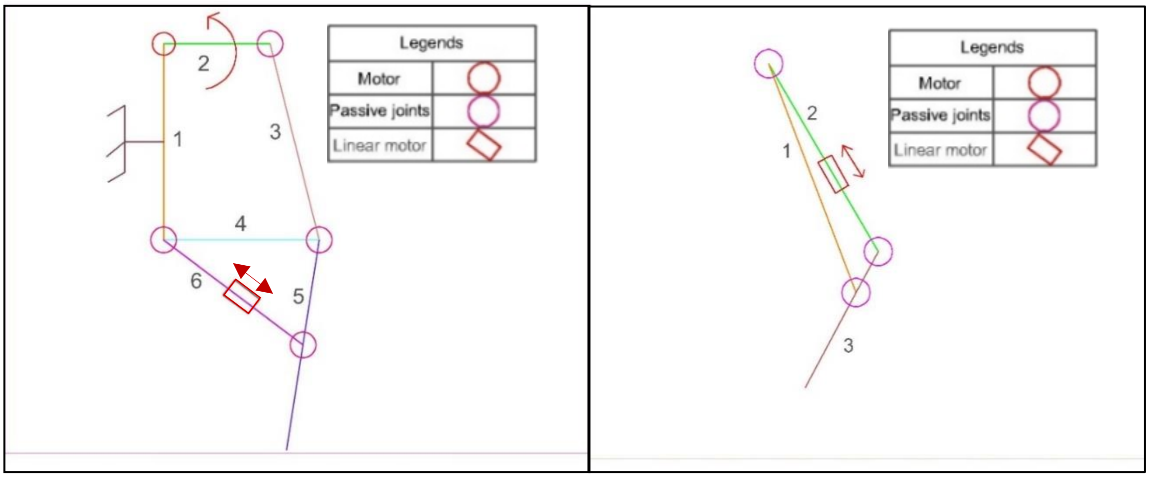

Figure 1: Alternative concept generations for the exoskeleton model

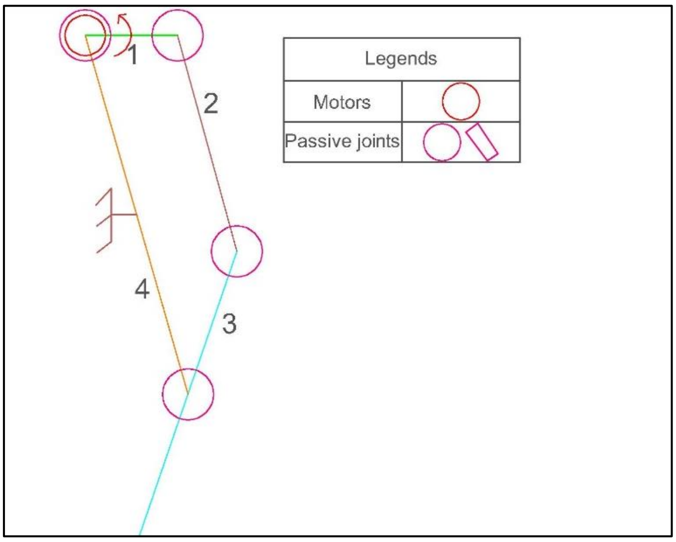

Figure 2: Selected conceptual design showing the linkage mechanism and joints

Figure 2 shows the selected conceptual design that is modelled as a modified 4-bar mechanism. The actuation of the linkage mechanism is generated at the crank arm (link 1). By using unequal lengths for the crank arm and rocker arm (link 3) it is possible to generate a larger torque at rocker arm using a lower torque at the crank arm, enabling the torque requirement of the linkage mechanism to be reduced.

Analysis using Grubler's equation (shown in equation 1) shows that the degrees of freedom of the proposed 4-bar mechanism are equal to 1 which means that the number of independent motions of the links in the mechanism is 1. Therefore, it is possible for the motion of the mechanism to be controlled through the motion of the crank arm.

$$F = 3(n-1) - 2l - h = 3(4-1) - (2 \times 4) - 0 = 1 \quad (1)$$

Where F is the number of degrees of freedom, n is the number of links, l is the number of lower pair joints and h is the number of higher pair joints.

The lengths of the linkage parts are determined to ensure that the exoskeleton is able to replicate the height and movement of the human user. Therefore, the selected link lengths must be assessed comparatively to the height of a user. The selected exoskeleton link lengths are shown in Table 1.

Table 1: The required lengths for the links of the exoskeleton to match the human user's height

Hip-knee length male (cm)	40.08
Knee-ankle length male (cm)	39.59
Total length at leg extended condition (cm)	79.67

The lengths of the proposed design of the 4-bar mechanism are selected accordingly and modelled in the ADAMS software interface as seen in the figure below: -

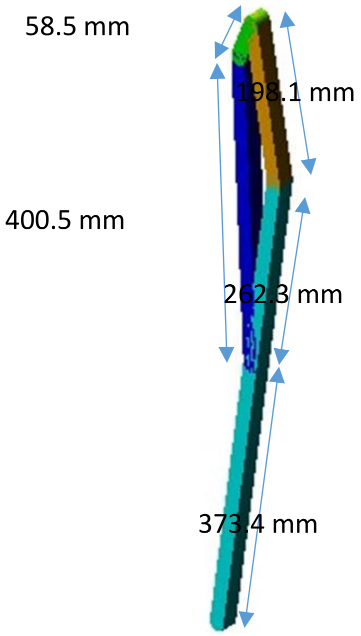

Figure 3: Lengths of the proposed 4-bar mechanism

2.2 Kinematic analysis

The kinematic analysis of the proposed 4-bar mechanism was conducted to generate the relationships between the angular velocities and angular displacements of the mechanism. This was done by applying motion functions to the crank arm (link 1) of the mechanism: -

$$STEP(time, 0.0, 0.0d, 5.0, 90.0d) + STEP(time, 10.0, 0.0d, 15.0, -90.0d) \quad (2)$$

This function induces a rotational displacement of 90° of the crank arm allowing the generation of the angular velocities of the rocker arm (link 3).

2.3 ADAMS modelling of exoskeleton and dynamic analysis

The exoskeleton design is then modelled in the ADAMS software by using the results of the kinematic analysis. The exoskeleton model is attached to a human body model to ensure that the system provides accurate results from the dynamic analysis. Actual weights of a 75kg user are added to the system to provide the necessary mass of the human body model. The weights of the individual body parts given to the human model are shown below in Table 2.

Table 2: The approximate weights of different body parts for a 75kg user

	Body segment % weight	**Appr. weight for 75kg user**
Thigh	9.88%	7.41kg
Shank	4.65%	3.49kg
Foot	1.45%	1.09kg
Torso	84.02%	63.01kg

The modified ADAMS model containing the exoskeleton and the human body model are shown in Figure 4. The 4-bar mechanism is connected to the human body model by the hip belt attachment which connects to the hip of the user.

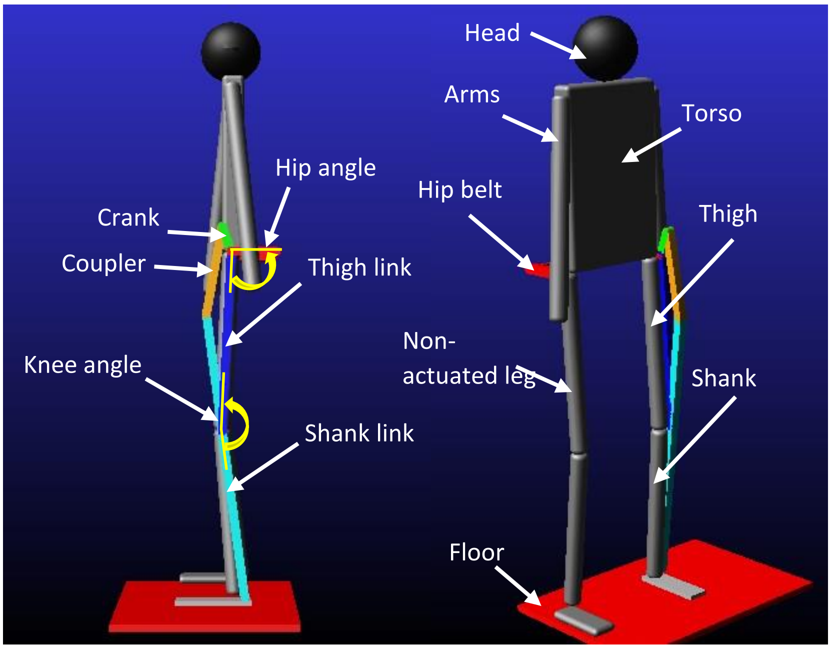

Figure 4: The modified ADAMS model showing the parts of the exoskeleton and the human body model with side-on and isometric views of the model

2.4 Addition of forces and ADAMS measures

The forces acting on the ADAMS model are defined within the system. The main forces applied to the system are the torque required at the crank arm (between the crank and hip belt), the reaction forces at the contact points of the users' legs and the floor, and the stability torque acting on the torso around a planar joint (within the sagittal plane) applied to the model to provide stability to the human body model. The movement of the model is restricted to the sagittal plane to balance the system.

The static and dynamic friction coefficients for the contact forces at the foot are given as 0.6 and 0.5 respectively. These values are consistent with the friction coefficients of the typical indoor floor surfaces. The crank torque and the stability torque of the model are controlled through feedback control loops using Simulink as the interface for the simulation. The addition of forces to the ADAMS model is shown in Figure 5.

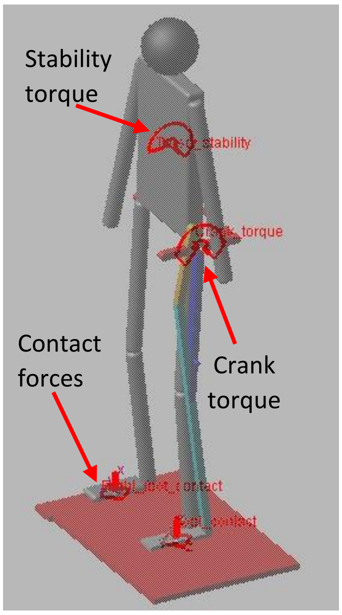

Figure 5: Figure showing the addition of forces to the model

Measures are introduced within the ADAMS software to measure the following parameters: -

1) Crank angular velocity
2) Shank link angular velocity
3) Thigh link angular velocity
4) Angular velocity of the torso

The angular velocities of the crank and the torso are used as inputs to the Simulink model. By generating the error signal between the measured values and the desired values the feedback loop of the Simulink model provides the signals to the PI controllers. The thigh and shank link angular velocities are used to generate the angular displacement of the knee and hip joints of the human body model.

2.5 Control system design

The control system design is done using Simulink software. The model uses feedback control to generate the required responses to the system. The feedback loops use the measures generated through the ADAMS dynamic model to generate the error signal between the desired and the actual signals as the inputs to the PID controllers. The error signal is used by the PI controllers to generate the required torque for the ADAMS dynamic model to function. The desired angular velocity of the crank is shown below in Figure 6.

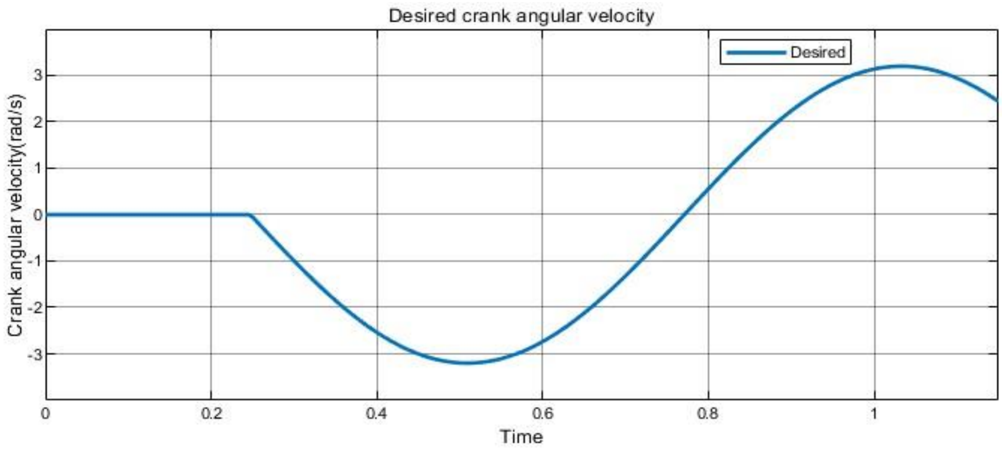

Figure 6: The desired crank angular velocity for the exoskeleton model

The PID control blocks used for the control of the crank and stability torque are shown in Figure 7.

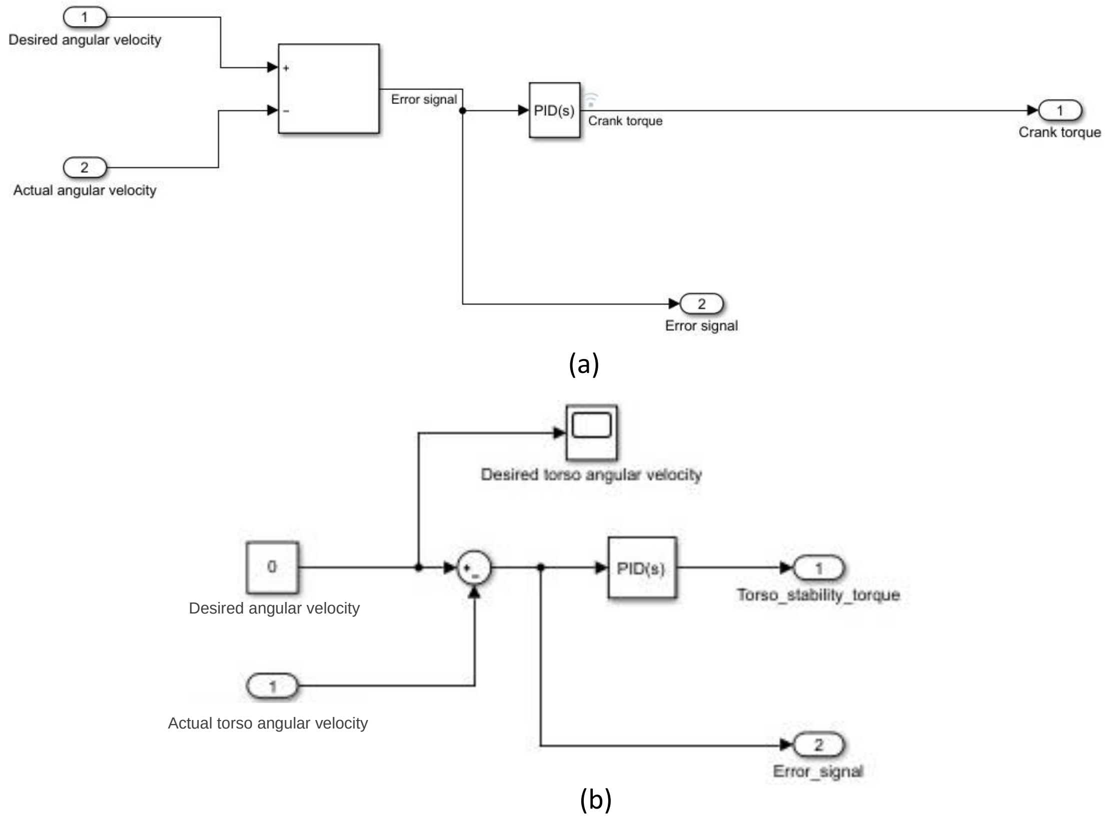

Figure 7: PID control blocks of the (a) crank torque and (b) the stability torque

2.6 Partial gravity balancing using spring

The proposed design of the 4-bar mechanism, although capable of actuating the exoskeleton, needs more support in order to actuate the hip joint of the user. Therefore, the addition of springs to balance the weight of the users' thighs is explored.

To calculate the parameters for the gravity balance springs, the mass of the body being considered must be assumed to act on one point and that the spring used is a zero-free length spring. This means that the spring exerts zero force when compressed fully. When these conditions are fulfilled (Arakelian, 2016) shows that by equaling the moment generated by the spring force and the moment due to the gravitational forces acting on the body, gravity compensation is possible. By using the following:-

$$Spring\ force\ (F_{spring}) = F_0 + k(l - l_0) \tag{3}$$

$$Potential\ energy\ (V) = mgs \sin\varphi \tag{4}$$

By balancing the moments of the spring force and the moment due to gravity,

$$V = \left(F_{spring} \times \frac{ar}{l}\right) \times \sin\varphi \tag{5}$$

Where *m* is the mass of the link, F_{spring} is the spring force, F_0 is the initial force of the spring, *s* is the length to the centre of mass of the link, *a* and *r* are the lengths from the pivot point to the spring connections and φ is the angle between the lengths *a* and *r*.

$$mgs = \left((F_0 + k(l - l_0)) \times \frac{ar}{l}\right) \tag{6}$$

For a zero-length spring, the relationship of $F_0 = kl_0$ is derived and therefore,

$$mgs = kl \times \frac{ar}{l} \tag{7}$$

$$k = \frac{mgs}{ar} \tag{8}$$

Equation 8 is used to calculate the spring stiffness required for a gravity balance spring. However, it must be noted that full gravity compensation is not required for this proposed design, therefore only partial gravity compensation is considered by reducing the mass to be lifted by the spring mechanism.

3 RESULTS AND DISCUSSION

The results of the kinematic analysis in Figure 8 show that the 4-bar actuates as expected, therefore the angular velocity of the rocker arm (shank link) would be proportional to the crank angular velocity. This means that due to the shank angular velocity being equal to the angular velocity of the knee joint, it is possible to control the knee joint motion by varying the crank angular velocity.

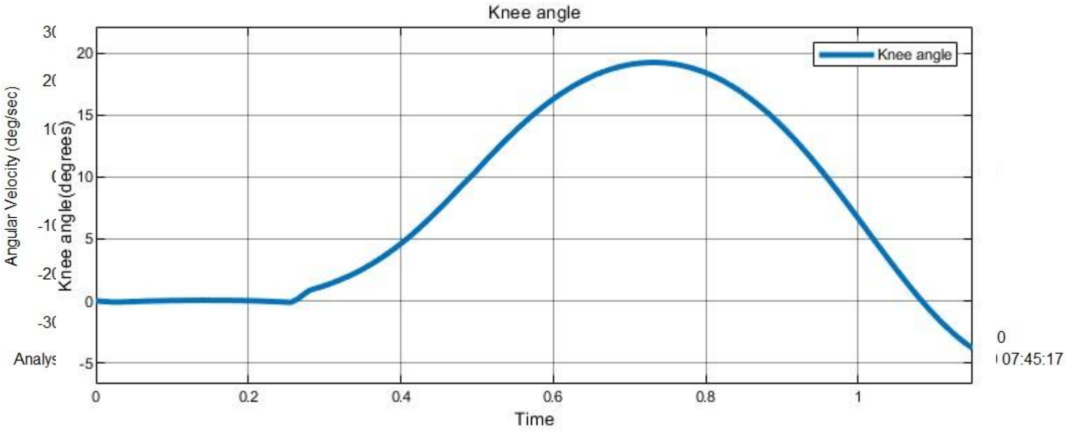

Figure 8: Rocker angular velocity with respect to the crank angular velocity

Initial dynamic analysis of the proposed design shows that the system is able to generate a good range of motion of the knee joint. The natural gait cycle of a person during the swing phase generates $30° - 50°$ of motion (Mentiplay et al. 2018), while the exoskeleton model induces a range of motion of around $20°$ By controlling the crank angular velocity, the required movement pattern of the knee joint was generated. The generated range of motion through Simulink is shown in Figure 10.

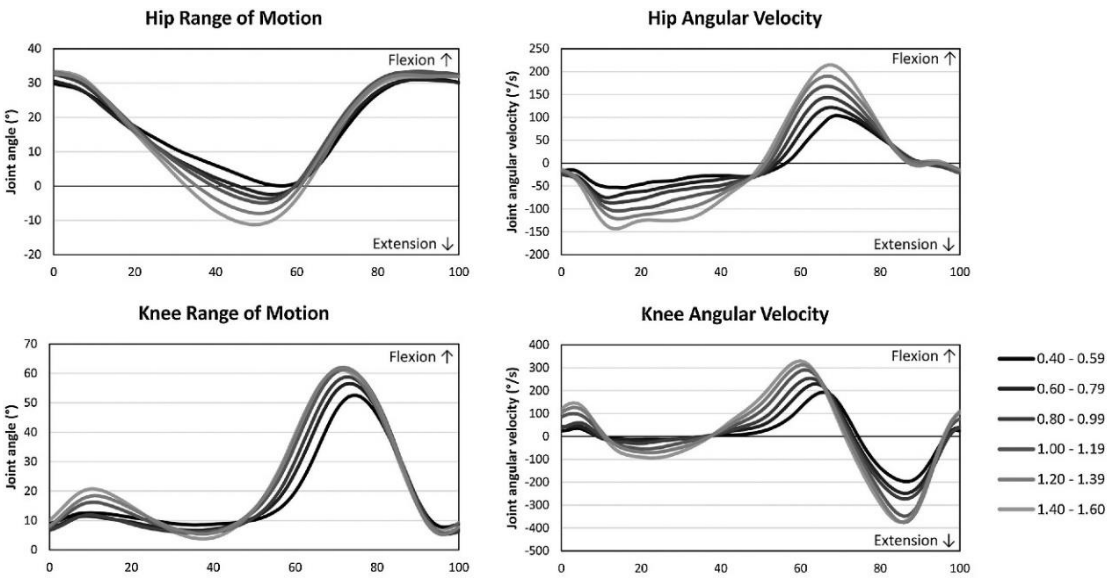

Figure 9: Natural range of motion of the human gait cycle (Mentiplay et al. 2018)

Figure 10: The knee range of motion generated through the initial simulations

However, the generated hip range of motion is relatively undesirable due to the dynamics of the proposed design as seen in Figure 11. This is due to the system inducing a negative range of motion of the hip joint as the exoskeleton rotates the user's legs backwards. The simulation results of the initial dynamic analysis shown in Figure 12 illustrates the dynamics of the simulation which show that the exoskeleton is able to generate the required motion of the knee joint at the halfway stage, even though at the simulation end, the user's legs move backwards due to the undesired hip range of motion.

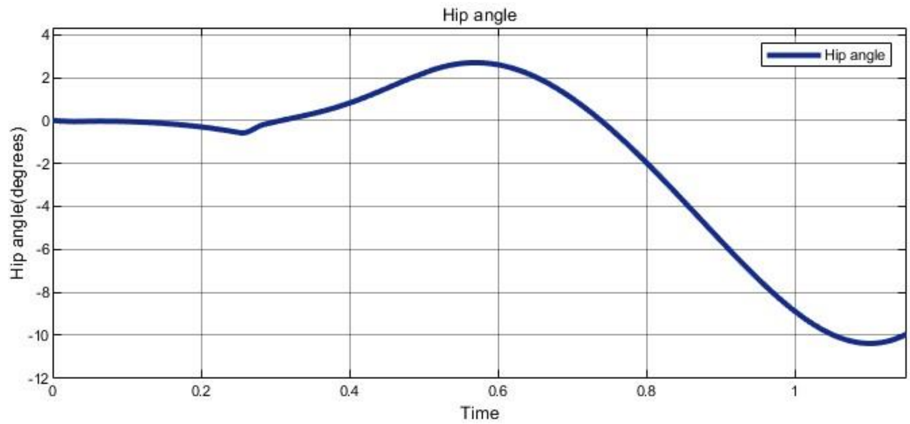

Figure 11: Undesired range of motion of the hip joint

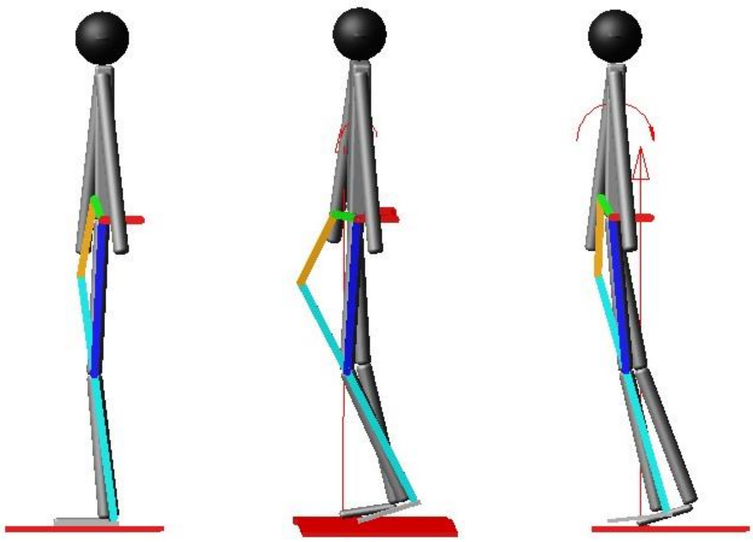

Figure 12: Simulation results of the initial dynamic analysis showing the movement pattern of the dynamic model at t = 0s, t = 0.6s, t = 1.2s respectively

The undesired range of motion of the hip joint results in the need for an alternative actuation method to generate the desired range of motion of the hip joint. Therefore, the gravity balance spring is added to aid the 4-bar mechanism in actuating the hip joint of the user.

The addition of the gravity balance spring improves the range of motion of the hip joint up to 10° as shown in Figure 13. However, in comparison to the natural gait cycle of $30° - 40°$, the exoskeleton hip range of motion needs improvement. The simulation results from the final dynamic analysis shown in Figure 14 incorporating the gravity balance spring, illustrate the improved motions generated through the addition of the springs. The improved model is able to produce better motion of the exoskeleton to allow for the desired motion at the end of the gait cycle by inducing better motion of the hip joint.

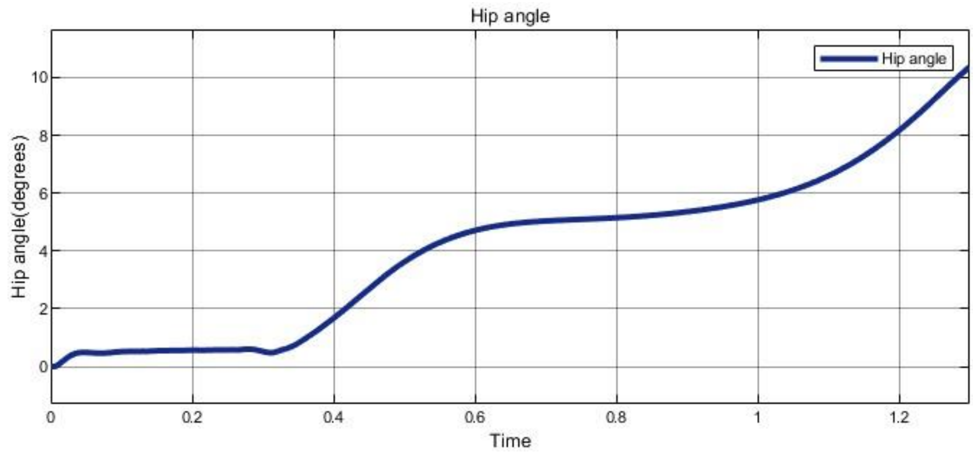

Figure 13: Improved hip range of motion through the addition of the spring mechanism

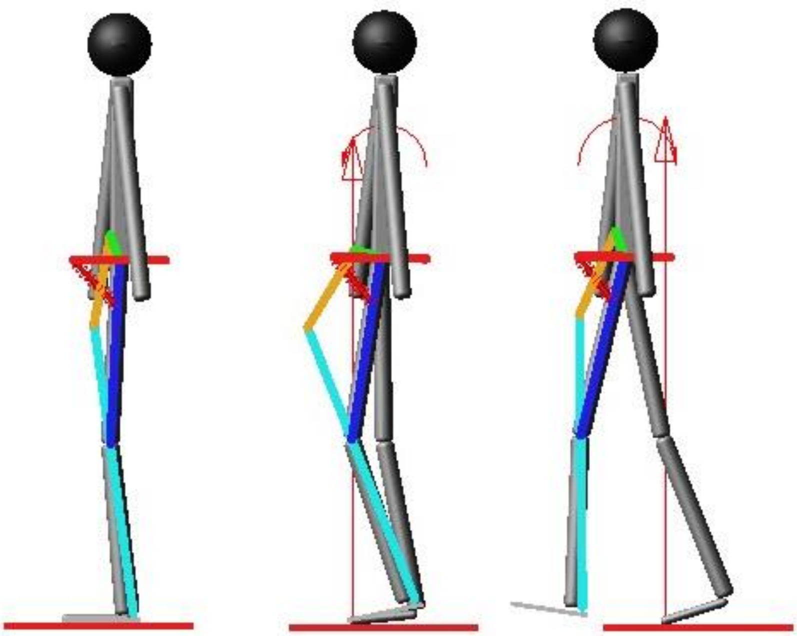

Figure 14: Simulation results of the final dynamic model with the addition of the gravity compensation springs at t = 0s, t = 0.6s, t = 1.2s respectively

The torque requirement at the crank is generated through the Simulink interface. Figure 15 shows the crank torque requirement of the exoskeleton.

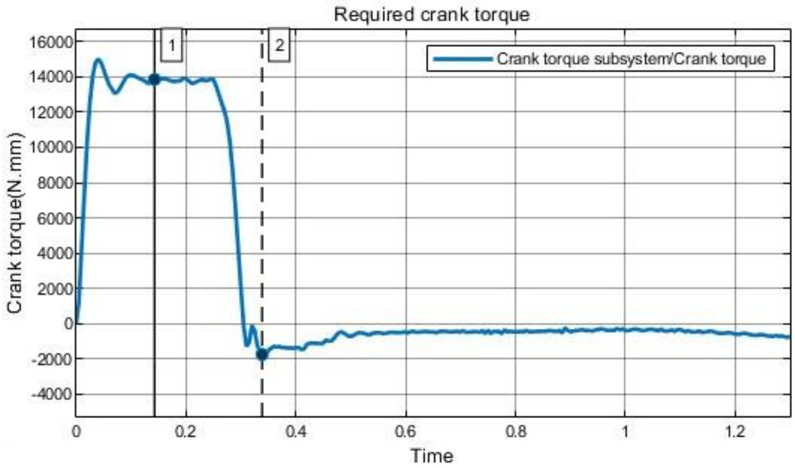

Figure 15: Crank torque required by the exoskeleton during the simulation

As seen in Figure 14, the crank torque peaks at around 14 Nm during the simulation. This torque is the torque required during the standing phase ($t = 0\ to\ t = 0.25s$) or when the exoskeleton bears the full weight of the model. During the swing phase ($t = 0.5s\ to\ t = 1.3s$) or during the flexion and extension of the knee joint, the crank torque can be seen to be below 1 Nm.

The reaction forces on the hip joint of the human body model are calculated to measure the impact of the exoskeleton on the users' body. The generated forces are shown in Figure 16. Through analysis of the forces, it is evident that the reaction force on the hip is generated due to both the crank torque and the moment of the spring force.

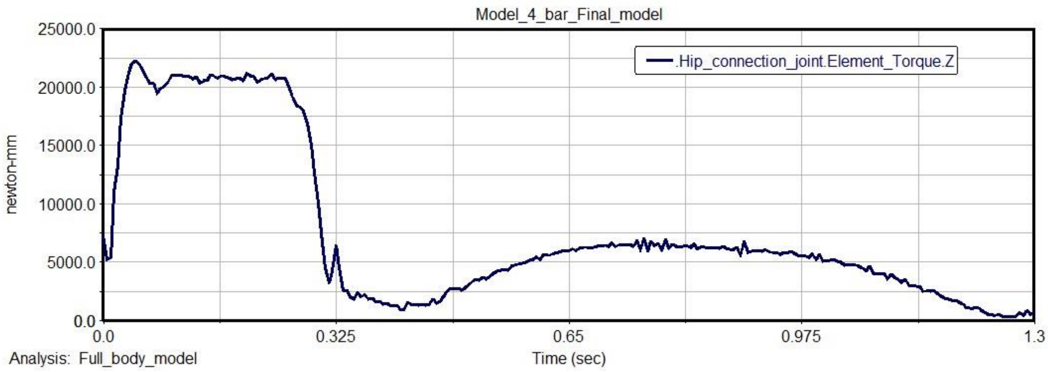

Figure 16: Reaction torque at the hip joint of the user due to the exoskeleton

4 CONCLUSIONS

The objective of this paper was to design and model an exoskeleton design that is cost effective and easy to manufacture but most importantly be usable for paralysis and stroke patients.

The exoskeleton is able to generate the required motions to allow the user to replicate the requirements of the walking cycle. The proposed design can actuate the knee joint of the user successfully using the single actuator of the 4-bar mechanism. The torque during the swing phase of the simulation is seen to be below 1 Nm without gravity compensation and under 2 Nm with gravity compensation. Improvements to the hip range of motion were done by the addition of the gravity balance springs to aid the motion of the exoskeleton. The spring mechanism was able to support the motion and enable the exoskeleton to generate a good range of motion for the hip joint. The overall system can induce a knee range of motion of 20 degrees and a hip range of motion of 10 degrees during the simulation. The feedback control system used for the system control was able to respond well to the requirements of the desired exoskeleton movements. The simplicity of the proposed 4-bar mechanism reduces the complexity of the exoskeleton to aid the manufacturing process.

Further improvements could be made to the exoskeleton design by optimizing the gravity balance spring mechanism and adding a framework to the exoskeleton to aid the user in withstanding the torque generated at the hip joint.

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