BMET5790 Introduction to Biomechatronics

Knee Exoskeleton Design Report

Method of Rehabilitation for Children with Cerebral Palsy

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1 Background

1.1 Cerebral Palsy

Cerebral Palsy is a neurological condition consequential of brain damage resulting in most commonly movement disability in children. The condition itself is one that has an array of effects in children inclusive of [2]:

- Movement and walking disabilities
- Speech difficulties
- Learning disabilities
- Cognitive impairments
- Hearing or vision loss
- Epilepsy
- Emotional and behavioural changes
- Spinal deformities
- Joint problems

This range of effects is primarily due to each child having an experience in relation to the condition. Cerebral Palsy is often a consequential condition of brain damage, however, there are various triggers of such damage [2]. The exact cause of the condition is not always determinable but often causes can include:

- Poor brain development - pre-birth
- Maternal infections and medical conditions
- Disruption of blood flow to the developing brain
- Genetic conditions
- Ingestion of toxins or drugs during pregnancy
- Damage to the head or skull during delivery
- Complications related to premature delivery

At this moment, there are four types of cerebral palsy known as **spastic, dyskinetic, ataxic, and mixed** [3].

1.1.1 Spastic

Spastic Cerebral Palsy results in increase muscle tone known as spasticity and consequently causes delayed developmental milestones for moving, abnormal movements, movement inhibition, stiff and spastic muscles, difficulties controlling muscle movement and difficulties moving from one position to another [4].

1.1.2 Dyskinetic

Dyskinetic Cerebral Palsy results in dystonia, repetitive and twisting motions, athetosis, writhing movements, chorea, unpredictable movements, poor posture, painful movements and difficulty swallowing or talking [4].

1.1.3 Ataxic

Ataxic Cerebral Palsy results in poor balance, limited coordination, tremors and shaky movements that are difficult to control [4].

1.1.4 Mixed

Mixed Cerebral Palsy results in two or three of the other types with the most common being Spastic-Dyskinetic Cerebral Palsy [4].

1.2 Knee Exoskeleton

A Knee Exoskeleton is a viable solution for children suffering with Cerebral Palsy, as it has the ability to relieve several conditions, such as Knee or Hip Subluxation, Dislocation, Spastic Movement, correct and limit or prevent Deformities, Low-Tone Pronation, Swing-Phase Inconsistency, Drop-Foot, Eversion and Inversion Turn Movments [5].

The Knee Exoskeleton Brace overall is a cost-effective and simple solution to the estimated two-thirds of children that suffer from Cerebral Palsy. This solution allows the children to gain the ability to walk and ambulate. Furthermore, the nature of Cerebral Palsy means that it establishes a gait for the affected muscles, joints and patterns of motion of the children [3]. The device itself therefore has the following benefits:

- Providing stable base movement
- Made of soft textures
- Made of common material
- Establish efficient gait
- Minimizing effects of spastic movement
- Reducing excessive energy exertion

- Creating an environment for children to repeat steps
- Adding transition from sitting and standing
- Reducing potential for accidents
- Increasing a child's ability to function physically and mentally
- Reducing or eliminating hip-knee hyperextension
- Strengthening weak muscles
- Controlling muscular imbalances
- Correcting poor skeletal alignment
- Preventing deformity
- Providing stable base for support

The key aspects that make this solution an appropriate one for that minimizes effects of the condition while also providing a low cost, easily repairable and maintainable solution that is quiet and efficient [6]. Therefore, the Knee Exoskeleton will assist children with mobility and assist them to maintain stability. Overall, allowing for a development not only physically but mentally with repeatable steps and motion assistance.

Key Terms - *Cerebral Palsy (CP), Ataxic, Dyskinetic, Spastic, Exoskeleton, Gait, Eversion, Inversion, Neurological, Deformities, Low-Tone Pronation, Drop-Foot.*

2 Introduction

The following document highlights the design of a Hinged Knee Brace Exoskeleton aimed toward the demographic of children suffering from Cerebral Palsy. The design encompasses a range of specifications to allow for assistance to the child with the motions of sitting, standing and walking and transition between them. Therefore, the overall design will assist and address the issue of motion being prohibited and limited with children that suffer from the condition.

2.1 Problem Identification

The specifications of this design include the Knee Exoskeleton to be focused on the use for children of various ages and is capable of providing sit-to-stand and walking assistance. Furthermore, the design has the following primary requirements:

2.2 Specifications

- Based on existing hinged knee brace - minimise the amount of development
- Adaptable to a wide range of children sizes
- Lightweight and completely portable
- Simple feedback and control mechanisms
- Quiet and efficient
- Powered by readily available energy sources
- Low maintenance
- Easy to repair
- Low cost (manufacturing and market)

2.3 Assumptions

The following section highlights the assumptions when designing the knee exoskeleton:

- Children of various ages (5 to 12)
- Contains electrical and mechanical components or integration in the design
- Low maintenance and repair considerations

3 Analysis

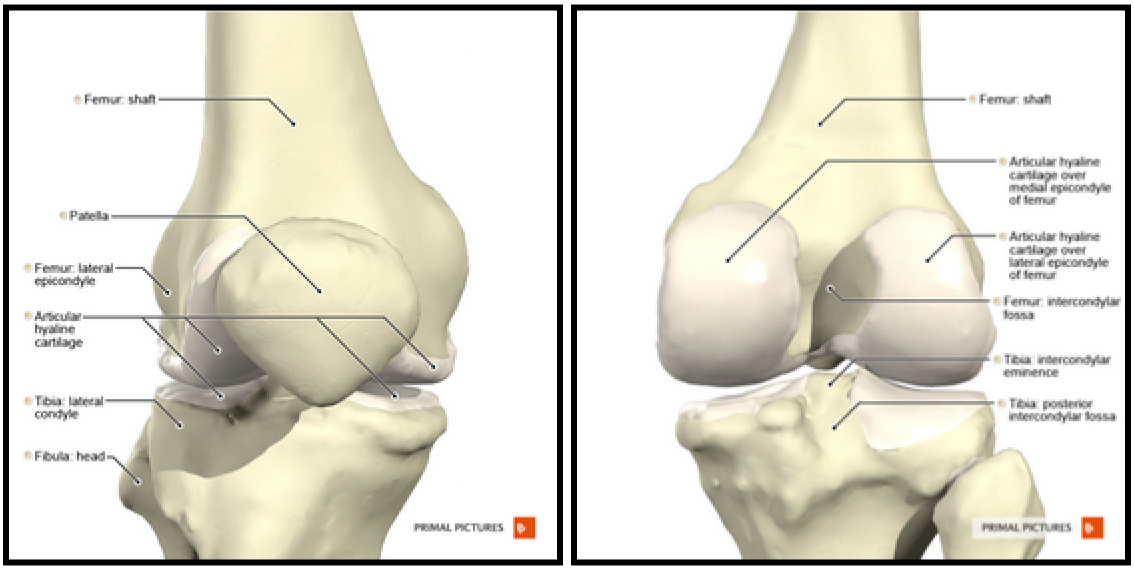
3.1 Knee Physiology

The knee is an integral component to providing support for a range of human movement. It serves a function in which the knee allows the leg to bend (flexion) and straighten (extension), as well as rotate the ankle inwards (internal rotation) or outwards (external rotation) [7]. The knee joint is formed by two small bones which allow for a synovial hinge articulation at the intersection of the distal component of the femur (thigh bone) and the proximal tibia (shin bone). This is supplemented with an extensive network of ligaments and muscles [8].

The fibula is a small bone that runs alongside the tibia, providing stability and assisting with rotation of the ankle, and is supported by various tendons which prevent it from dislocation [9]. The anterior cruciate and posterior cruciate ligaments prevent the femur and tibia from sliding front and back, respectively, relative to each other - whereas the medial (inner) and lateral (outer) collateral ligaments prevent the femur and tibia from sliding left and right [9][7]. The medial and lateral collateral ligaments also act as a horse-shoe shaped lining, providing shock absorption and reduction in friction, distributing lubrication in the form of bursae (fluid-filled sacs).

The patella is attached to the patella tendon, a muscle which holds it to the upper portion of the tibia, and to the quadriceps tendon, holding it to the front of the thigh. The patella primarily functions as a pulley for the quadriceps, allowing for more effective knee flexion and quadriceps strength [9]. It serves a protective function, covering the aforementioned network of ligaments. To hold the patella towards the centre, the trochlea groove notches it in place.

Figure 1: Anatomical Diagram of the Knee



3.2 Knee Mechanics in Cerebral Palsy

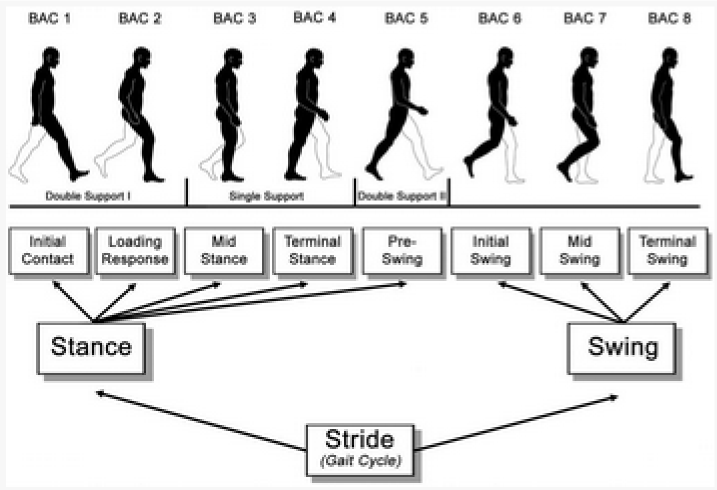
The knee acts as a hinge joint, meaning its primary movement is flexion-extension. As per any synovial joint, the femur's articular surface rolls and glides over the tibial surface. The impetus for this movement is provided by a contraction force in the quadriceps. This means that in the case of someone with CP, spasms in this muscle would not allow for proper articulation of the knee's hinge mechanism [10].

4 Quantitative Analysis & Design

4.1 Gait Cycle Angles

Gait refers to the cyclic coordination of the movements of arms and legs to propel the torso forward. The arms and legs move in such a way that weight is transferred and caught between the left and right side of the body. As such, a gait cycle is one full stride composed of various phases in stance and swing [11].

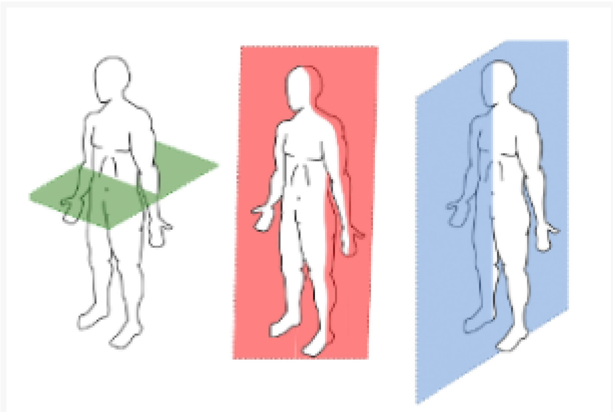
Figure 2: Typical phases throughout the gait cycle



A gait cycle analysis allows for a multi-perspective understanding of an individual's locomotion throughout a single stride. It follows two periods, which can be broken down into different phases and plotted against various spatial and temporal parameters. In this way, one is able to comprehensively understand the mechanisms of control and diagnose movement disorders [12].

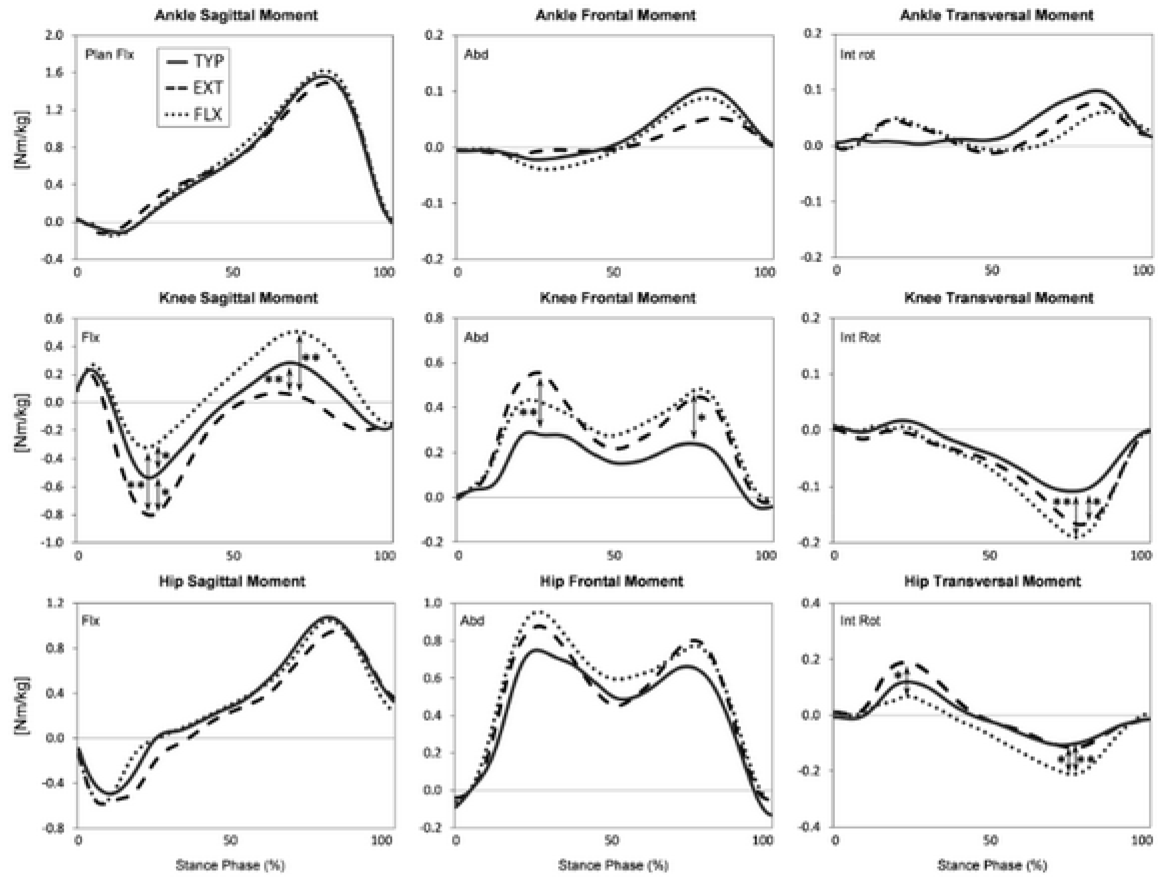
In the context of anatomy, different perspectives are referred to as planes. Figure 4.1 depicts the various anatomical planes.

Figure 3: Anatomical Planes - transverse (green), frontal (red), and sagittal (blue)



A gait cycle analysis takes into consideration various metrics such as the force, torque, and moment across all three axes. In the case of the knee, the hinge joint allows operation primarily across the sagittal axis - however, the aforementioned abduction/adduction and internal/external rotations must also be considered.

Figure 4: Typical gait graph, multiple planes and metrics



4.2 Force & Torque

The forces and torques that are acting upon the knee of a child aging from five to twelve years therefore, is a range of differing values. In order to ensure safety of the system, the values that are being utilized for the structural design includes the average forces in X,Y and Z directions of the knee upon the maximum aged child. These values must be utilized to allow support for children that fit within this age group. The resulting force and torque values are calculated through a set of equations derived via the creation of a FBD.

The forces resulting from knee movement is separable through firstly looking at the unbraced knee during: [13]

Figure 5: Free body diagram - frame of reference

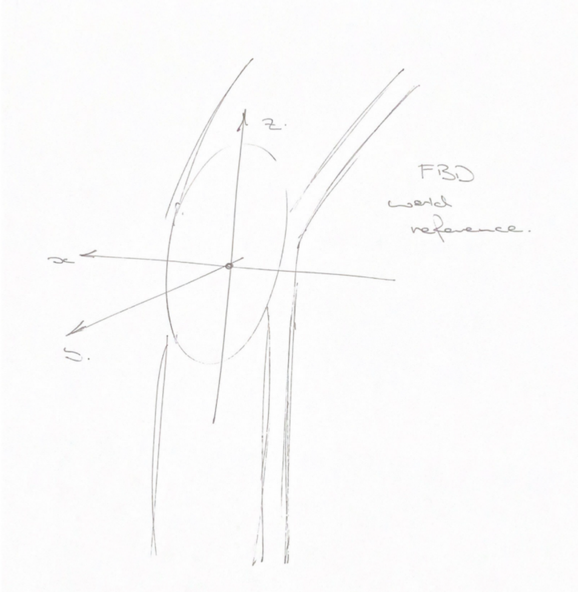
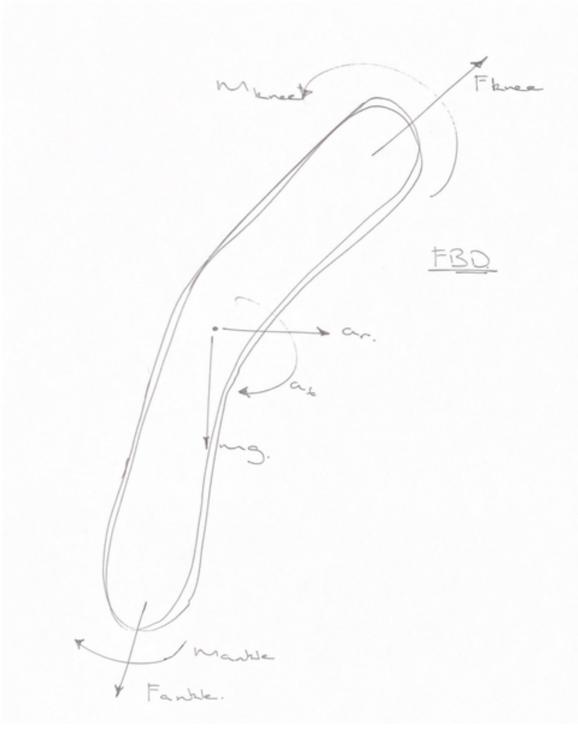


Figure 6: Free body diagram



These peak values of force and torque in X, Y and Z directions are an average and normalized using body weight of each subject. Therefore, the tibial forces are hence, multiplied by the body weight of the child. The following table demonstrates the final force values for walking,

Force (N/kg)		
Direction	Sitting	Standing
Fx	0 to 0.5	0 to 1
Fy	0 to 1	0 to 1.5
Fz	-2	10 to -12

Table 1: Force p/kg w.r.t each Directional Axis

Torque (Nm/kg)		
Direction	Sitting	Standing
Fx	0 to 0.2	1 to 1.5
Fy	-1 to -0.5	-3 to -3.5
Fz	-0.2 to 0	0.1 to 0.2

Table 2: Torque p/kg w.r.t each Directional Axis

Activity	**Peak Tibial Forces (N/kg)**	**Notes**
Walking	2.5 to 2.8	Floor
Sitting	0.4 to 0.5	Chair
Standing	1.25	Floor

Table 3: Force acting on Tibia p/kg for each movement

sitting, standing and transitions [14].

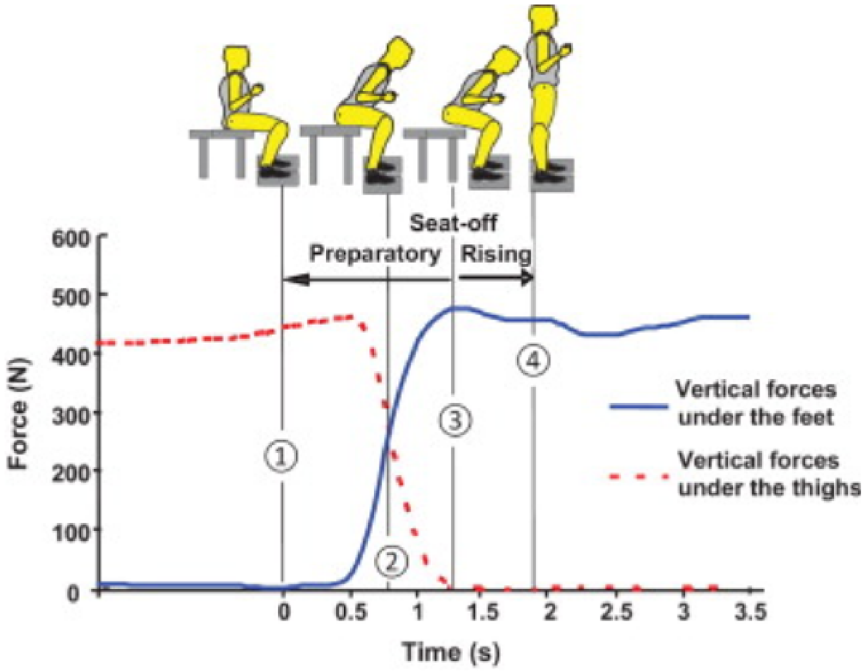
The sitting to standing transition simply incorporates the peak values of sitting and standing individually, therefore, there is no requirement to analyse the transition aside from looking at the peak when performing a sit to stand motion or pivot [14]. The overall range of the force for sitting to standing is: 400-500N.The investigation should therefore, incorporate into design the capability to reduce the 500N force that is endured by the tibia when performing this motion [15].

4.3 Energy

The following section highlights investigation and analysis of energy related values for the sit-to-stand transition and movement. Furthermore, this section highlights any considerations related to the energy:

Note: This is the specific researched values relating to the energy exerted by CP patients of younger age (specifically children of maximum age 12).

Figure 7: Example analysis for children movement



The overall time elapsed for the sit to stand transition is **about one second** to keep consistency and control on the energy expended for the movements surrounding the knee. In conclusion or comparison with other data from younger children that do not suffer from CP require three to five times less the energy.

Activity	Value	Units
Standing	0.05	kcal/min
Sit to Stand	0.18 to 0.78	kcal/min
Sitting	0.02	kcal/min
Walking	0.21	kcal/min

Table 4: Energy exerted by child for each movement

4.4 Resulting Base Design

Therefore, with considerations of the loads and increased energy that is acting on the knee for most young CP patients, the following is the primary mechanical and basis of the design of the knee brace. From this point we will add additional electronic components to create a final mechatronic design. Through research of a selection of mechanical components and techniques that can reduce load around a pivotal point, the most effective components to be implemented include a spring system and specific selection of the material.

4.5 Liquid Spring Hinge

The implementation of a spring loaded hinge to the knee brace which allows for the overall load acting on the knee to be reduced and support the leg through specifically sitting to stand movements and while walking or standing. The springs absorb the force specifically going through the knee similar to that of a shock absorber, hence, reducing pressure on the joint. The energy compressed in the springs allows for the energy of the patient excerting when walking and performing compressive movements of the knee to be reduced [16].

Figure 8: Liquid Spring Hinge Example

The liquid spring hinge is a viable design to implement on the side of the as they are more powerful with greater results than solid spring knee braces - absorb more force or body weight and assist in knee extension movement with reduced pressure on the joint. The component is adjustable from most suppliers - adjusted through the spring hinge.

Through CAD force analysis acting on a pre-made knee brace model, there are many areas of the knee brace that can be rid of material. Furthermore, this has no adverse effects on displacement or stress when the knee is undergoing load. Moreover, the reduction of material reduces cost, weight and allows for better comfort as the knee and surrounding region is more breathable.

Figure 9: Overall Current Design (Front View)

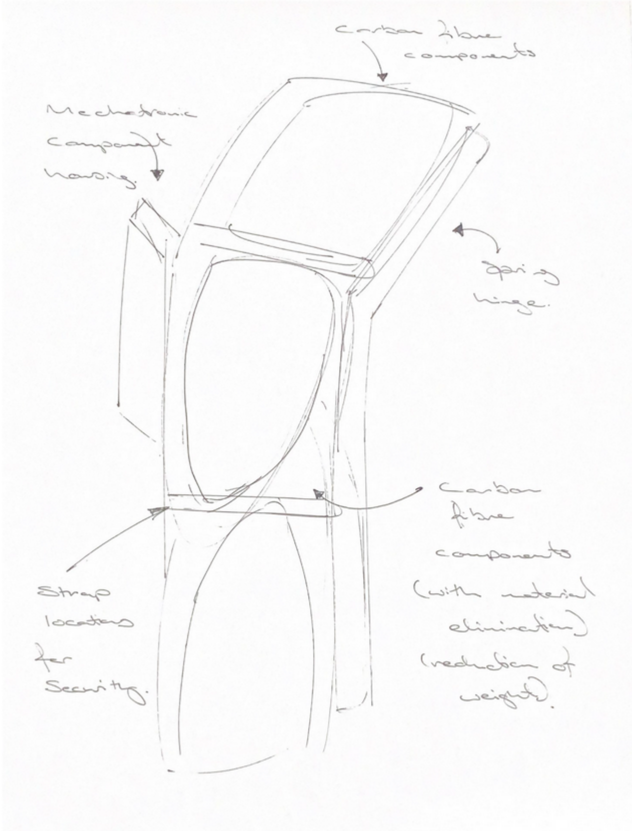
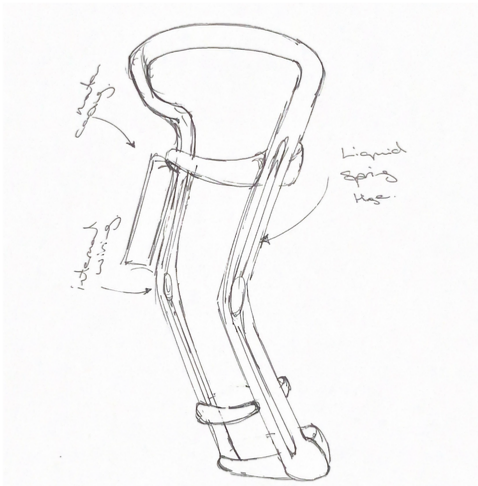


Figure 10: Final Design of Knee Brace (Simplified View w/Frontal View Emphasis)



4.6 Material Selection

The frame material selection has a profound effect on the overall performance of the robotic exoskeleton. As the requirements of the exoskeleton are derived from the supporting forces provided to the tibia and femur, the frame's material must be selected such that it minimally impedes this transfer of force. The weight of the exoskeleton is an important consideration, and by proximity, the material selected for the exoskeleton must be light.

Most exoskeletons employ some variant of aluminium alloy (typically 707) [6], as the concept of an exoskeleton for human use is associated with rigidity. The objective is to reduce the weight of the design as much as possible, but allow for physiological forces to be conveyed between the struts. Titanium is likely to be an ideal material given its ratio of strength to weight, providing a yield strength to density of 60% greater than aluminium [17]. However, titanium is also substantially more expensive (at about $40 per kilogram), and its bioactive properties - which sees its natural use in implants - is not required in this case.

An alternative material that can be used is carbon-fibre, among other fiber-reinforced plastics. Carbon fibre does not permanently deform below its ultimate tensile strength, so it effectively does not have a yield strength (or yield strength to density) [18]. Although it does not provide the rigid ceiling of titanium, carbon fibre is substantially lighter. To combat titanium's greatest weakness of cost, carbon fibre costs around $21.5 per kilogram. Overall, it is logical to use carbon fibre as the shell for the CP knee exoskeleton, especially in the case of a design that only actuates in a single degree of freedom.

5 Exoskeleton - Mechatronic Component Design

5.1 Literature Review

The knee joint, as the main joint responsible for movement of one's lower limbs, is the most vulnerable and susceptible joint in the human body. In the process of human movement, the knee joint plays the main role of supporting the body weight, assisting lower limb swing and absorbing impact. Children with CP show quantifiable lower limb weakness and joint muscle imbalance [19]. As an effective and conservative physical therapy, knee assistance devices have continuously attracted scientific research in the past few years. The knee exoskeleton is one such popular option, with many innovations in various designs.

5.1.1 Backdrivable Electro-Hydrostatic Actuator

In 2010, Hiroshi Kaminaga et al. [20] proposed adding backdrivable (reversible) capabilities to the knee exoskeleton. In order to present such capability, they designed a hydraulic actuator based on Electro-Hydrostatic Actuation (EHA) [20]. It is modified and designed based on the lower body strength (measured in annual torque and required force) of elderly patients. The overall design logic is worthy of reference. One can also change the reference object to children with CP, adding backdrivable functionality to the exoskeleton to reduce spasticity and overall pain.

5.1.2 A Pediatric Knee Exoskeleton with Real-Time Adaptive Control:

Adding a micro-controller to provide maximum continuous assistance torque of 15Nm at the velocity of 360°/s is shown to improve child's comfort [1]. This can be taken into consideration to design a complete control system.

5.1.3 A Self-Adjusting Knee Exoskeleton for Robot-Assisted Treatment of Knee Injuries:

In this research, a new type of active device for robot-assisted rehabilitation is proposed [21]. It is improved according to the 3RRP Mechanism and can adapt to the transitional movement and rotation of the knee joint, thereby achieving a perfect match between the human body's joint axis and the equipment axis. With automatic adjustment of its joint axis, the proposed equipment not only guarantees ergonomics and comfort during the entire treatment process, but also expands the usable range of motion of the knee joint. In addition, the adjustable function significantly shortens the setup time required to connect the patient to the exoskeleton, allowing the patient to spend more time effectively on exercise instead of making adjustments [21]. The special feature of this research is that it can support passive translational movement of the knee joint and independent active control of these degrees of freedom. In addition, the author also introduced the implementation details of a prototype that has a compact design and combines the power of three actuators to achieve high rotational torque [21].

5.1.4 An Adaptive Knee Joint Exoskeleton based on Biological Geometries

To obtain the influence of different exoskeleton designs on the internal joint forces, a dynamic model of the interaction between the knee joint and the double-link exoskeleton was proposed [22]. This model has a significant influence on the joint strength of the knee joint by the closed kinematic chain of the leg and exoskeleton. Based on the knowledge of knee joint kinematics, the adaptive knee joint exoskeleton is designed through combining different moving parts (such as pins, sliders, and cam profiles) [22]. This design has the potential to eliminate the negative effects associated with the closed leg/exoskeleton kinematic chain on the human knee. The flexion motion of the artificial human knee joint was studies to compare the performance of five exoskeleton designs to those without an exoskeleton. The results obtained using the dynamic model (based on the characteristics of the knee joint) are very similar to the experiment [22].

5.2 Mechatronic Design

The mechatronics design of the exoskeleton used in this study was inspired by previous research [1] and was optimised according to the research's findings.

5.2.1 Design Requirements

Firstly, the mechatronic equipment should provide a maximum continuous assistance torque of 15Nm at a speed of 360° per second. The source of this target value is from the child's

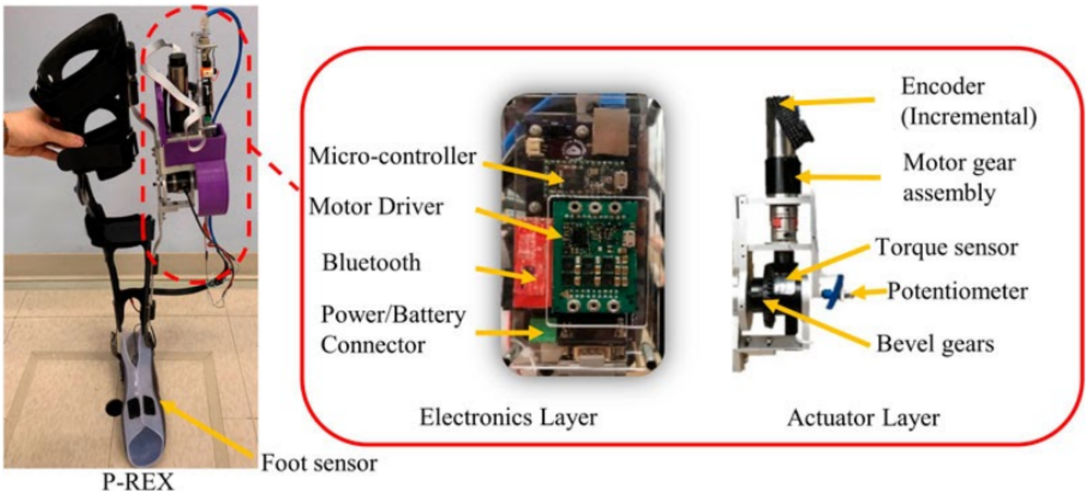
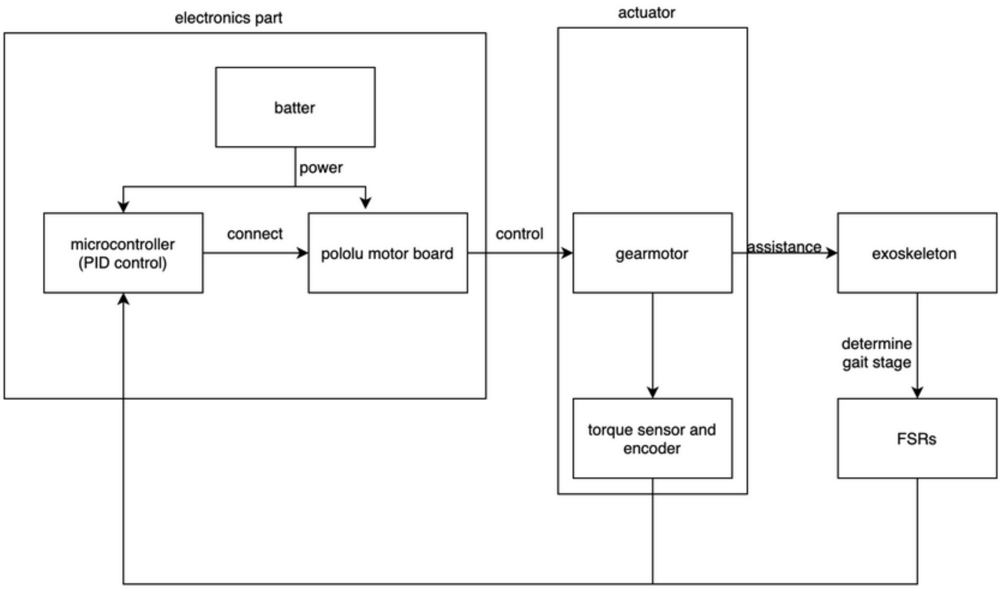


Figure 11: The overall mechatronic design-actuator assembly and embedded electronic control system mounted on the lateral side of the leg [1].

Figure 12: Block diagram of the new mechatronic component's design logic



peak swing knee extension speed of 350° per second [1], and studies have also noted that usually the developing child's maximum torque exerted upon their knees during walking is 1Nm/kg. The actuator of the equipment should be able to provide a maximum of the 50%

of the maximum knee torque for a child weighing about 30 kilograms.

The whole device adopts a right-angle transmission device design, allowing the actuator to be installed on the outer thigh (figure 5.2) to facilitate putting on and taking off the exoskeleton, and to reduce the possibility of restricting the extension of the knee joint on the plane of travel. Furthermore, it adopts a modular design, making assembly and repair expedient.

5.2.2 Design Logic

From the block diagram in figure 5.2, the logic of the mechatronic operations are visible. Using FSR sensors mounted on the bottom of the exoskeleton or feet to derive the gait state, as well as a torque sensor result with a microcontroller to the PID control, provides suitable torque for the current gait state. The specific assembly method is based on the model in figure 5.2, but the selected components have been updated and changed to have a smaller volume (including reduced gear, through the use of a gear motor).

5.2.3 Gear Motor Selection

Figure 13: SGMADA PG36555 Gear Motor

A 12-volt, DC gear-brushed gear motor is selected (SGMADA part number PG36555). This DC motor is reliable, has low vibration, can be sterilized, and provides a long shelf life. It is capable of 15N/m stall torque at 12V, satisfactory to the target. The motor output shaft is fitted with a multi-stage planetary gearbox, having a selective multiplicity reduction ratio. This selection can also reduce the weight and cost whilst simultaneously providing ample torque.

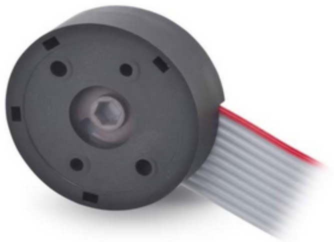
5.2.4 Torque Sensor Selection

As shown in figure 5.2, the built-in reaction torque sensor is installed on the knee shaft between the gear and the calf attachment. This sensor is selected as the PC900 series rotary drive torque, providing industry-grade precision.

5.2.5 Encoder Selection

The MR series encoder, with a capacity of 500 counts per revolution, is used to meet the specified resolution of 0.1° angular position and 0.1°/second speed [1]

Figure 14: The MR Series Encoder



5.2.6 Micro-controller and Motor Board Selection

The microcontroller component is used to power the actuator and provide the feedback control for the motor. The microcontroller's model is within the PIC18 family, and the motor's model is a dual VNH3SP30 motor driver board (Pololu 707), as shown in figure 5.2.6. This circuit provides an ST VNH3SP30 fully integrated H-bridge motor driver for each motor. Each motor driver is controlled by a TTL-level PWM input that controls the motor speed and two TTL logic inputs (INA and INB) that control the direction of drive or stopping. Thus, the motor speed is controlled by the microcontroller and the motor driver board.

5.2.7 Foot Sensor (FSRs)

The foot sensors (FSRs) are used to get the pressure of the foot and judge whether users are standing or swinging, which is important for control. To provide accurate pressure data at a resonable cost, the Foot Film Pressure Sensor High Accuracy IP67 is selected.

5.2.8 Battery Package

Power is provided to the actuator and electronics by a Zippy Flightmax 4.2 A-hr Lithium Iron Phosphate (LiFePO4) battery. This battery integrates four series-connected cells to produce a nominal voltage of 12.8V. The battery is rated for a constant discharge rate of 30° (126A) and a maximum charge rate of 2°C (8.4A). This battery is rechargeable, which could reduce the cost. The battery is secured to the bottom plate (electronic layer) using velcro.

Figure 15: Pololu 707 Motor Driver

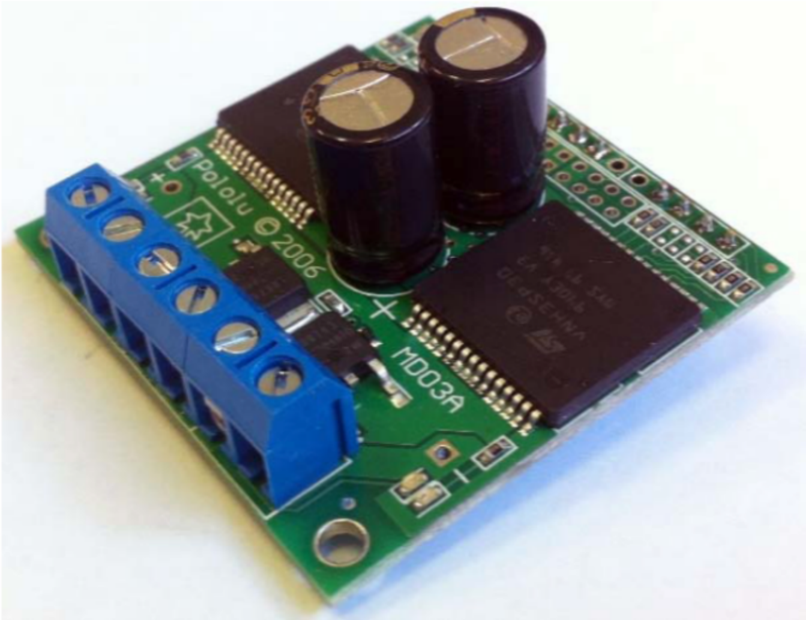
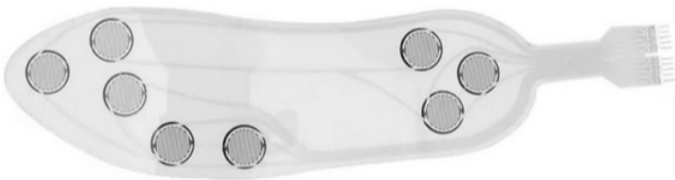


Figure 16: Foot Sensor



6 Feedback & Control

6.1 States

In the reference control design notes [1], the classification of the entire control system is very detailed, which is of great significance for the overall control thinking design. At first, the aim of the control part is to regulate the knee extension assistance during stand and swing stages in real time. The supervisory level consists of the finite state machine (FSM) [1], and splits the gait cycle into five states - early stance, mid-stance, late stance, early swing, and late swing. The states are divided using the predefined threshold to compare with the pressure from the feet (standing or swinging) and torque value from the torque sensor (vel). The threshold can be determined by weight and torque speed of the CP patient.

Figure 17: Battery

6.2 Mode Selection

There are three modes for each state, which are constant torque control, impedance control, and adaptive control [1]. The first adaptive control mode is currently still in the experimental stage, whereas for the swing stage, the effect of assistance is not tested [1]. As for the constant torque control, a comprehensive method gives constant assistance torque for all states, but contradicts with the initial design idea (real-time tuning of the assistance force). Thus, the second mode (impedance control) is the most suitable mode to select, as it relates the assistance torque with the knee angle and tunes the assistance according to such. In the mechatronic design, this mode will be considered as the only mode in which the user can achieve real-time knee assistance functionality.

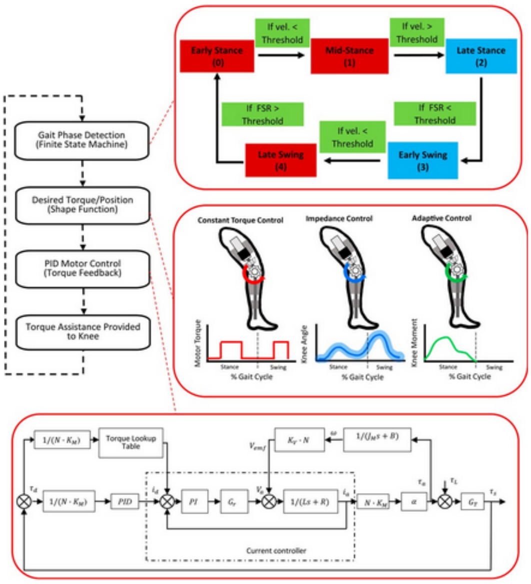
6.3 Control Loop

The PID control is used to reduce the unmodelled disturbances from the motor or external environment [1]. In the controller design, the PID control is combined with feedforward compensation. The feedforward can torque reach to the target level faster, and the feedback control could minimize the difference between actual torque and target torque [1]. Furthermore, A pre-defined torque reference lookup table is needed to determine the required torque for different states. The table should be based on multiple testing. Thus, the overall logic is to use the PID control to minimize the external environment error, and then compare this value to torque lookup table to get the required torque value - then use feedforward control to make the motor reach desired values and loop again.

7 Discussion

The mechanical analysis of the knee brace that has been designed overall effectively reduces and assists the patient with all movements defined in this investigation. The design is compact in comparison to most marketed knee braces that incorporate springs, however, the design has an increased weight in comparison to other solutions on the market. This is primarily due to the addition of the mechatronic or electronic motor components contained

Figure 18: The hierarchical real-time control loop running on the embedded microcontroller during exoskeleton operation (including five finite states, three models, and a close control loop [1])



on the exterior of the knee. To compensate for this carbon fibre has been selected as the primary material of the structure and exterior. Furthermore, the electronic component can have the wiring concealed within the hull of the design and motor driver utilizing smaller integrated boards to reduce the weight and space occupied by the electrical components.

7.1 Cost

7.1.1 Cost of Mechanical Components

Part Name	Quantity (per Brace)	Price ($, total)
Carbon Fibre	2sqm	190
Padding Material	1.5sqm	45
Velco Straps	3	50
Liquid Spring Hinge	2	300-400
Knee Frame Manufacturing	1	250

Table 5: Related costs of mechanical components

The total cost is approximately about $885 for the mechanical components. This assumes carbon fibre's market rate of $21.50/square metre, however moulding and manipulation costs make up the remaining $47/square metre. This is also the case for upholstery.

7.1.2 Cost of Mechatronic Components

The total cost is about $511 for the mechatronic components.

Part Name	Quantity (per Brace)	Price ($, total)
Gearmotor	1	12
Torque Sensor	1	80
Encoder	1	150
Battery Package	1	80
Micro-controller	1	12
Motor Board	1	63
FSRs (Foot Sensors)	2	2×62
External Bracket	2	2×20

Table 6: Related costs of mechatronic components

The total cost of the mechanical component in conjunction with the mechatronic component would incur a total cost of approximately $1500 to $2000 depending primarily on the manufacturing of the knee brace frame with the carbon fibre material and spring hinges implemented or assembled.

7.2 Maintenance & Repair Considerations

The whole mechatronic component uses a parts assembly, meaning that it is easy to disassemble and replace the parts. However, each part within the mechatronic assembly has typical working life limitations. For example, the gear motor can work only to 1200 hours maximum. Lubricant can be added to the motor as a daily maintenance measure. If the assistance sensor from the motor is delayed or functioning abnormally, the sensors can be tested. If they are broken, they can be disassembled and replaced.

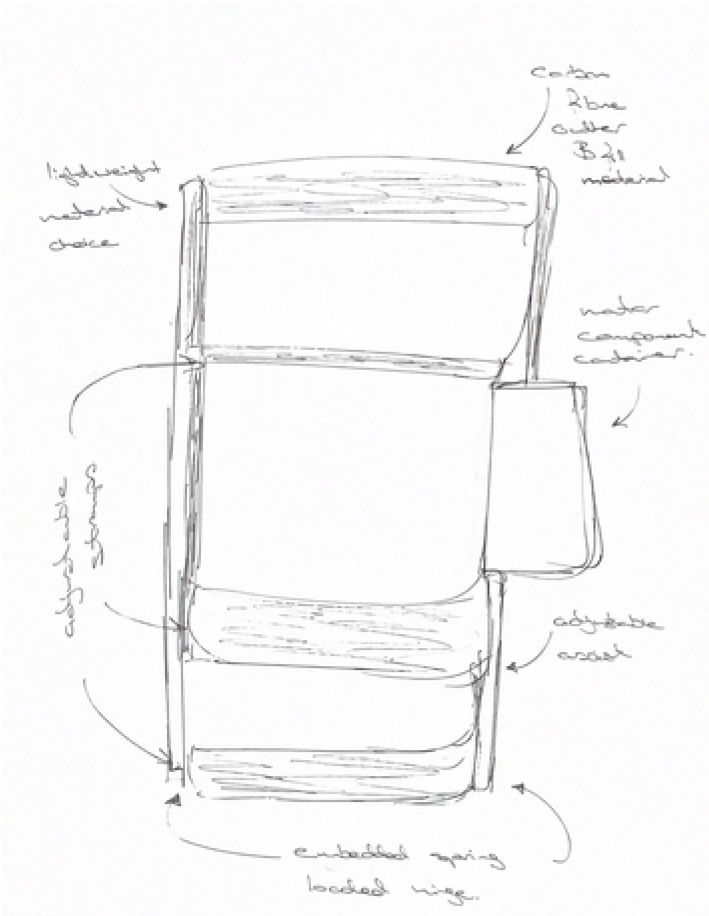
The majority of materials do not require regular maintenance and reparations aside from damages whereby they can be mended through carbon fibre and padded material. The overall adjustments and repairs to the knee brace can be performed using basic tools or even more technical kits such as Pod Knee Brace Replacement Ligament Sets. Overall, reparations costs can range from the total cost of any of the components above, otherwise, kits and tools can range from 40 to 100 dollars.

8 Conclusion

The whole mechatronic part is a combination of the methods mentioned by previous researchers and some of my research on hardware. The goal to be achieved is clear, and I also believe that real-time knee assistance can be achieved. However, some of the parameter settings should be supported by actual experiments. Although the entire control and electrical theory is straightforward and practical , the final version and threshold setting still requires a lot of testing and real-patient experiments to reach the perfect iteration.

The knee brace that has been developed during this investigation therefore, incorporates primarily mechatronic design and approach. To conclude, the solution incorporates a variety of modifications to current knee braces on the market and combines many different solutions of knee braces to independent problems that CP patients face. The design includes from the physics and mechanical analysis a carbon fibre outer structure with a padded interior along with velcro straps on the outer surrounding the knee and upper thigh and upper shin. Finally, each side of the brace will include an integrated liquid spring system to reduce energy exertion from the patient and absorb forces. With further development from the mechatronic investigation section, the unit can be contained (with interior wiring through the carbon fibre hull) into a small compact structure on the exterior.

Figure 19: Final Design of Knee Brace (Front View)

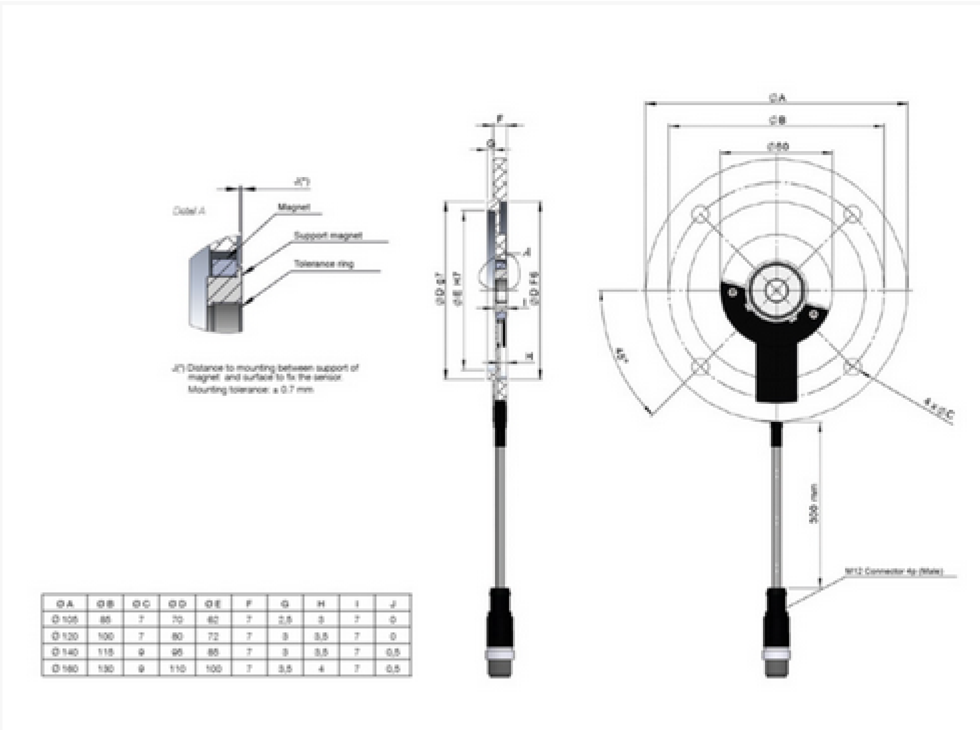


9 Appendix

Model Number	PG36555
Type	Brushed gear motor
Stall torque	15N/m
No load speed	10000Rpm
Life	1200 hrs

Table 7: Hardware datasheet - SGMADA Part Number PG36555 12V DC gear brushed gear motor

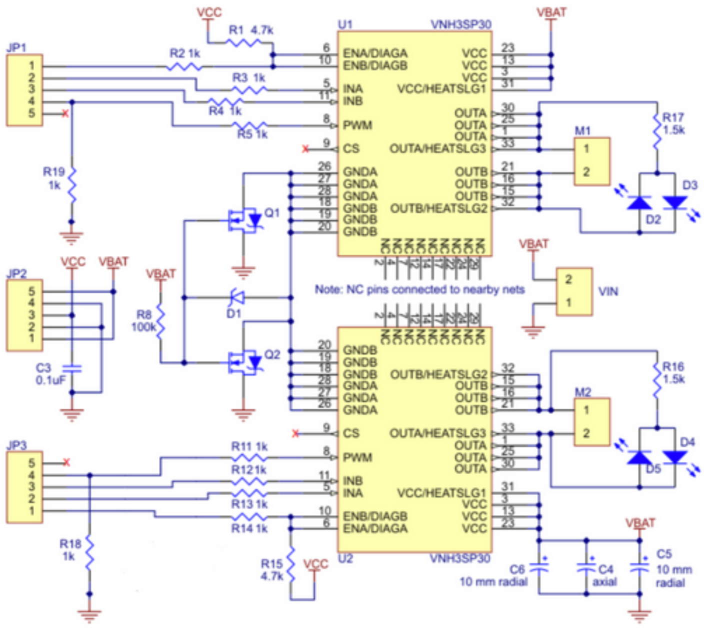
Figure 20: MR-Encoder



Capacity	4200mAh
Voltage	4S2P / 13.2V
Discharge	30C constant / 60C burst
Weight	535g (including wire, plug, and case)
Dimensions	138mm , 45mm , 47mm
Balance Plug	JST-XH
Discharge Plug	5.5mm Bullet Connector

Table 8: Battery Pack Specification

Figure 21: Pololu 707 Dual VNH3SP30 Motor Driver schematic



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