Design and Fabrication of a Semi-Passive Upper Body Orthotic Device for Recovery of Function/Mobility and Rehabilitation of Human Spine

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My work is dedicated to my parents and siblings whose tremendous support and cooperation led me to this wonderful accomplishment.

Abstract

An upper body exoskeleton system or an orthotic device intended for patients with severe spinal cord injury (quadriplegia) which allows them with the ability to sit from a lying posture and vice versa on their own. This system composes of a chain of plate's mechanism covering the back of patient/subject at different regions of the spine with 3 passive hinge joints between each plate junction (linkage between two plates). The axis of rotation for one joint is parallel to medio-lateral axis and two of the others have axis parallel to the antero-posterior axis. This provides each plate with a motion capability of 2 Degree of freedom (DOF) i.e. rotation along medio-lateral axis and restricted rotation at antero-posterior axis. Outfitting the system to give patient an ability to sit from a laying posture and conversely while providing spine with the flexibility (frontal plane) in the form of rotation along sagittal axis.

The actuation of the system is achieved through using two similar lead screw mechanisms positioned lateral to the abovementioned passive assembly. These lead screw mechanisms comprise of a rotating bar each with an axis parallel and lateral to the joint axis between hip section and L1(first lumber vertebra) that is actuated by the lead screw. Each bar is connected to the lateral ends of the top plate mechanically.

Hence, the passive assembly is within a rectangular frame consisting of active components (actuation mechanism) of the assembly located at the side of the frame, parallel to the torso. The top side of the fame is an angle iron (ALU) connecting the two actuator sides to fixate the frame while the lower side is made of aluminum plate which pivots the passive assembly at the bottom most plate of the passive assembly, through a hinge joint.

Key words: *Orthotics, Quadriplegia, Biomechanical, Kinesiology, Spinal curvatures*

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List of Abbreviations

DOF	Degree of Freedom
S.S	Stainless Steel
M.S	Mild Steel
Alu	Aluminum
COM	Center of mass

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CHAPTER 1 Introduction

Many people around the world lose the ability to control their body. There are a number of reasons for such disabilities. It can happen due to injury to the spinal cord resulting from accident, it can be due to stroke and trauma resulting in the paralysis of human body. Paralysis is the inability to control human body.

The level of such disabilities varies from patient to patient and depends on the intensity of the injury taken by the human body [1]. Mostly because of such injuries human lose the ability to control certain regions/segments of the body which in turn hinders in the interaction of such patients with the environment and limits the mobility of such patients [2].

Patients who lose the ability to control their lower extremities are referred to as paraplegic patients and such loss of mobility is mostly due to breakage in the lower segment of the human spinal column. As a result, patients facing such issues are unable to walk around or make any sort of locomotion with their lower limbs [3].

Similarly, patients who become unable to control all four limbs of theirs are termed as quadriplegic patients. The interaction of such patients with the environment or the surrounding is immensely limited as such patients cannot make any sort of defined motion with their upper and lower body [3]. Such disabilities make the life of such patients meaningless as whole of their body is out of their control and it puts a psychological bar on them which is not good for their growth and hampers in personal development [4]. Many to all patients facing quadriplegia are not content with their life and are immersed in negative feelings pertaining to their low mobility and due to the dependency on others for even basic tasks as sitting up right [5].

The focus of this research is on quadriplegic patients who lose the ability to control their torso (upper body and lower body alike) and are even unable to sit from a laying posture and vice versa. The set research objective for this thesis was to design and a fabricate a biomechanical palliating system or an orthotic device which allows for such bed ridden patients to sit from a laying posture and vice versa while keeping in mind the inherent biomechanics of the human spine. So that such posture change can be achieved in a comfortable manner for the patient. One other research objective which was set was to make the device user friendly. Which meant that the device needed to be safe to operate, easy to equip, not hard to operate and provides for comfortable postural changes in accordance with the kinesiology of spine and normality of the spinal curvature [6].

One important research objective which was set was to make the device portable. Which means that we wanted to come up with a device which was light weight, and which could be shifted from one place to another without requiring more than one person. We also considered portability in the broader sense that this device could be placed on any bed and turn that into an adjustable bed so that the device has the capability to equip any bed with allowance for such motions (postural change between laying posture and sitting upright posture). So, the device would work as an adds on to any bed and can be placed on any bed and still achieve the fundamental research objective: a biomechanical palliating system or an orthotic device which allows for patients to sit from a lying posture and vice versa.

CHAPTER 2 Literature Review

2.1 Paralysis

Paralysis is such a condition of human body in which there occurs loss of muscle function in a certain part of one's body. The soul reason behind it is issues in the communication between the brain and the muscles. Paralysis has different types and classifications [10]. There is complete paralysis and a partial one. It can also occur on one side or both sides of human body. It can be focused on a specific area or it can be wide spread (in some regions of human body). Paralysis in the lower body, including both the legs is called Paraplegia whereas paralysis of all the four limbs is known as quadriplegia. The most common reasons for paralysis are strokes or injuries to the body such as an injury to the spinal cord [8]. Apart from these there are other reasons for a paralyzed body such as nerve disorders and autoimmune diseases.

According to a study carried out in the year 2013 (by Christopher and Dana Reeve foundation on prevalence of paralysis in the United states of America) out of 50 people living 1 person faces the problem of paralysis [9]. Which adds up to a staggering number of approximately 5.4 million people. This number is equal to the combined population of Philadelphia, L.A, Washington D.C [7].

All of us have someone – a family member, a friend, a neighbor or a colleague who is living with some form of paralysis.

Some important findings of the study suggested:

- Paralysis is widespread. 5,357,970 people are reported to be facing some form of paralysis.
- Leading cause for paralysis is Stroke then spinal cord injuries and then multiple sclerosis
- Only 15.5% of paralysis patients are employed. Whereas 41.8% paralysis patients indicated that they could not work.

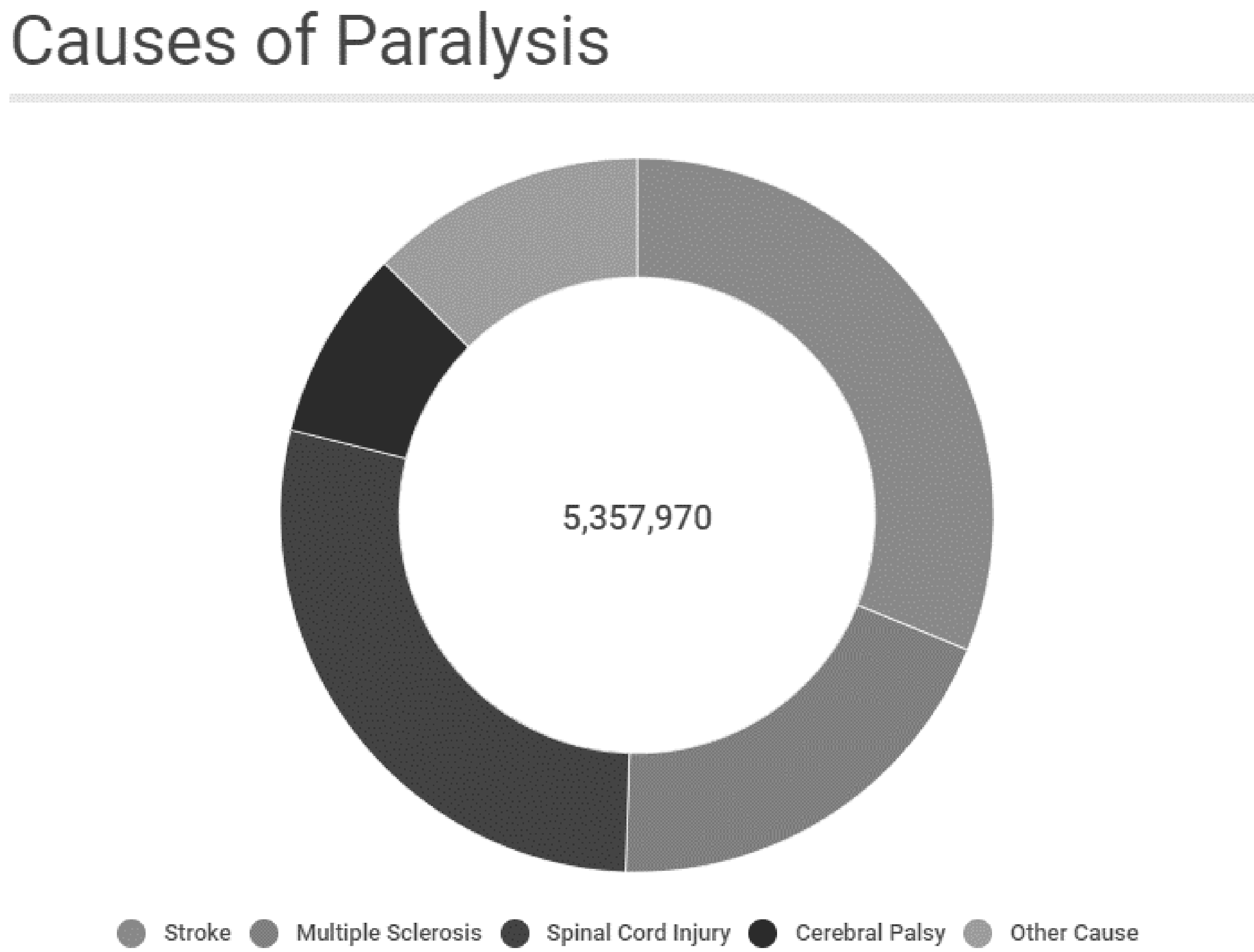


Figure 1 Causes of Paralysis

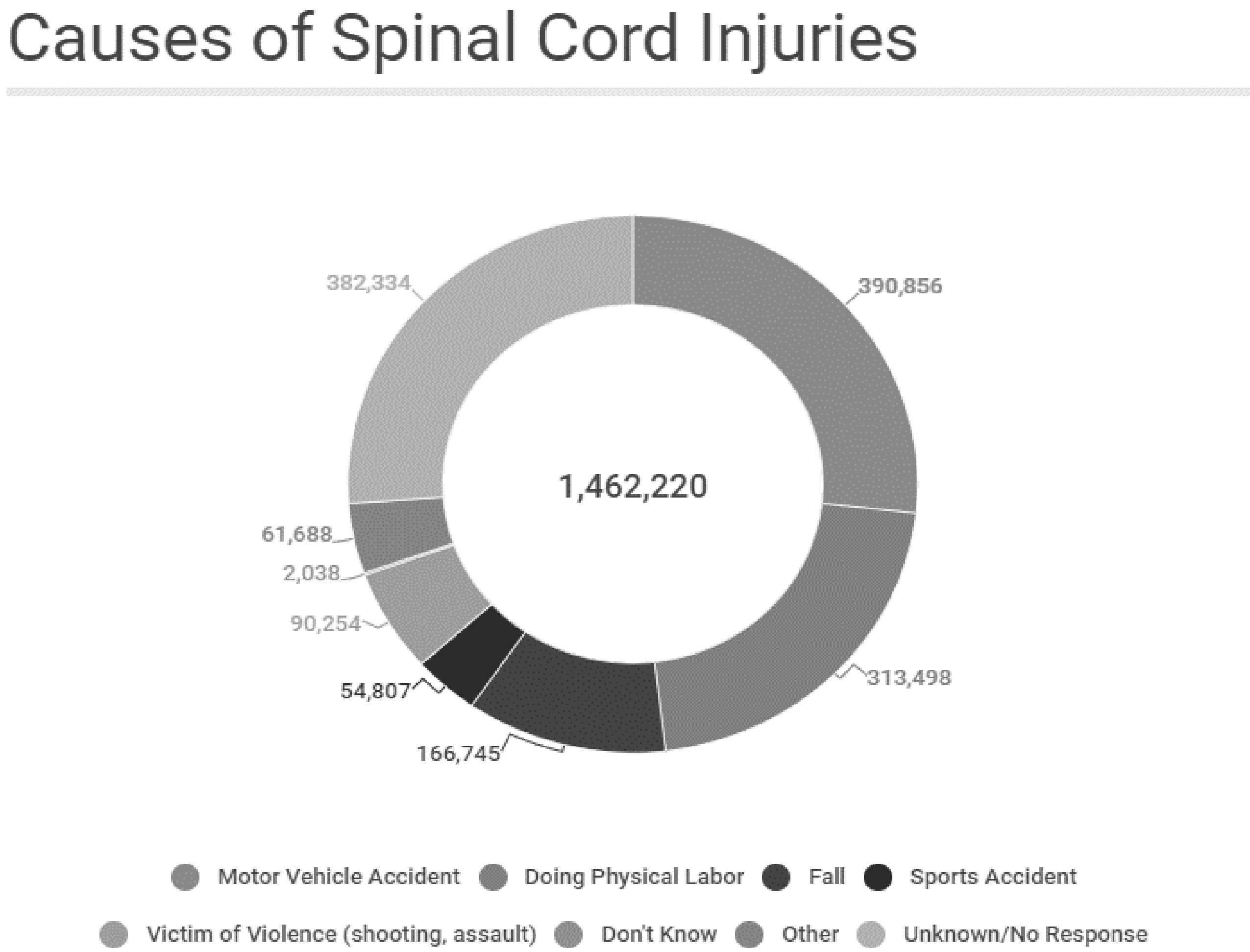


Figure 2 Causes Of Spinal Cord Injuries

This statistic pertains to any type and form of paralysis face by the human body.

It can be seen that paralysis is a widespread issue faced by the world today and acts as a hindrance in the interaction of patients with their surroundings [11]. So, there is a need for doing something to better the interaction of such people with the world or their surroundings [12].

For the scope of this research we choose patients with quadriplegia. These are such patients who lose the ability to control all four extremities and their torso as well. As for all paralysis, stroke is the number one reason behind such paralysis then is Spinal cord injuries. If we consider spinal cord injuries. The degree of impairment changes with the change in the segment of spinal column which is injured. The following figure will better clarify the context of the above asserted assertion.

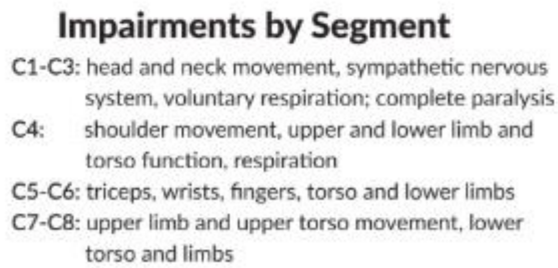


Figure 3 Impairments By Segment

As you can see people facing injury at segment c7 to c8 lose control of all limbs and the upper torso [14]. So, such patients lose the ability to even achieve a basic motor task of sitting from a laying posture. We set our goal for this research to come up with a biomechanical palliating system which can provide recovery of function of human spine for such a person to sit from a laying posture and vice versa while mimicking the dynamically changing spinal curvatures while achieving such postural changes [13].

Previous Work has been done in the past for the betterment in the interaction with the environment of people who face paralysis and also to provide comfort to the human body. One of the major domains for such solutions is Orthotics.

2.2 Orthotics

Orthotics has a Greek origin and the literal meaning suggests it to mean “to straighten” or “to align”. It is a specialty within the biomedical engineering field and pertains to the designing, manufacturing and application of orthosis. An orthotic device or an orthosis is an externally applied device (to the human body) which is used for modification of structural or functional characteristics of the skeletal system or the neuromuscular [16].

An orthotic device may have the following uses:

- Assistance in movement
- For Reduction of load bearing forces for any specific purpose
- Restrict motion in a certain direction
- To control/limit or mobilize/immobilize a joint, extremity or a body segment.
- To correct the shape of the functionality of a region of body

Orthotic solutions require a fused knowledge of Human anatomy, physiology, Pathophysiology, human biomechanics and engineering.

Currently there exist many orthotic solutions. There are current examples in the form of elbow orthoses, shoulder orthoses, orthosis for lower limbs and much more [15]. There are passive orthoses, semi passive orthosis and active orthoses also exist.

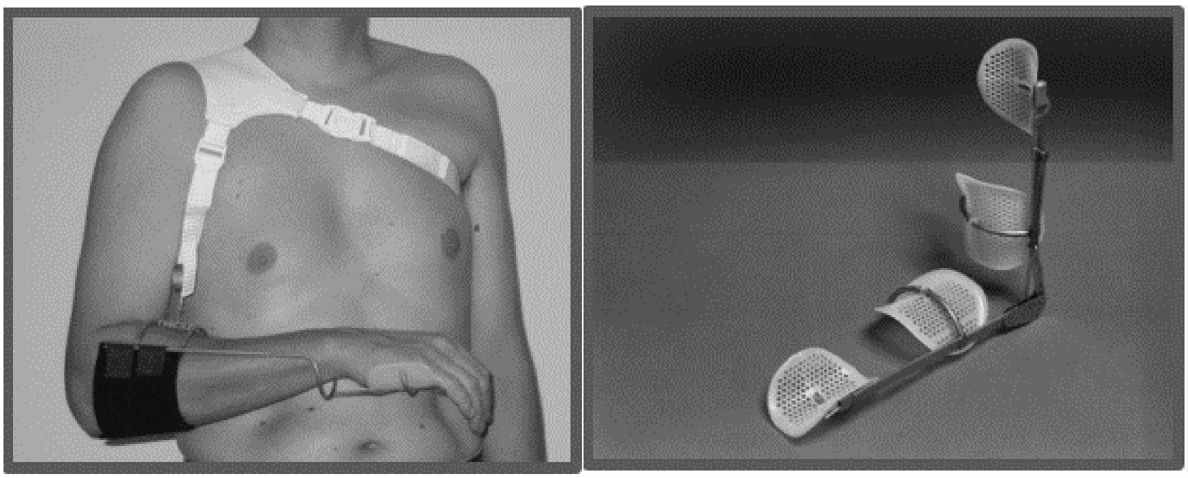


Figure 4(a) Passive Shoulder Orthosis (b) Passive Elbow Orthosis

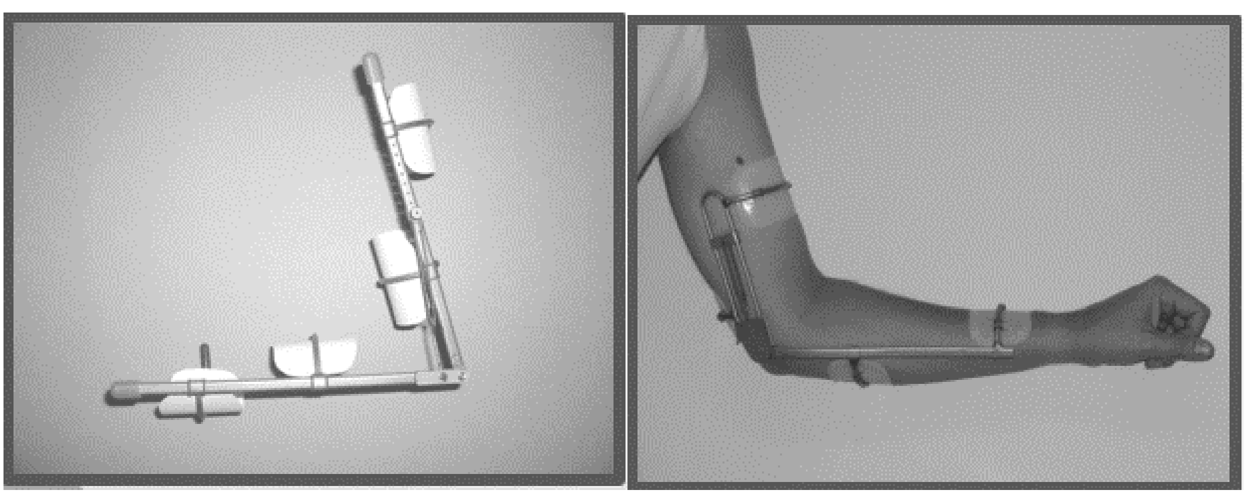


Figure 5 (a) Passive infant Elbow Orthosis (b) Semi Passive Elbow Orthosis

Figure 6 Spinal Orthosis to treat scoliosis

2.3 Other solutions

2.3.1 Adjustable beds

If we look specifically to our fundamental goal which was to make quadriplegic patients to sit from a laying posture and vice versa. Currently there is only one viable solution in the market which is in the form of adjustable beds [17].

An adjustable bed consists of a multi hinged lying surface which can provide different number of profiling/positions. Such beds can provide inclination to upper body and lower bodies independently [18]. Other more complex positions can also be achieved using these beds. Such beds have been in hospital use for a very long time but now are also being used in the home care domain for the last three or so decades. There are manual beds and then there are automatic beds available as well.

There are a few problems and limitations that we face with using adjustable beds which have been quoted below:

- It limits the use of mattress. Special mattresses need to be used with such beds which can allow for the necessary bends needed to achieve different profiling with the bed.
- These beds usually require a mechanism to provide the actuation and for that reason are usually very heavy and at the same time very large in size. This makes these beds poor at portability and usually requires installation charges as well along with the overall pricing of the bed.

- As far as the cost is concerned there exists a range depending on the brand/features provided. But on an average range such beds cost around 1200-2200. There are also others which cost more than 10000$ as well. So it can be seen that such beds can be quite expensive.
- The most important limitation that is faced by such beds is that even though they provide for postural changes and different profiling of the human body. They do not allow for mimicking of the biomechanics of the spine while doing so.

  A few examples of adjustable beds have been shown below:

Figure 7 Examples of Adjustable Beds

To better understand the limitation pertaining to the inability of such beds to mimic or follow the inherent biomechanics of the spine refer to the figure below.

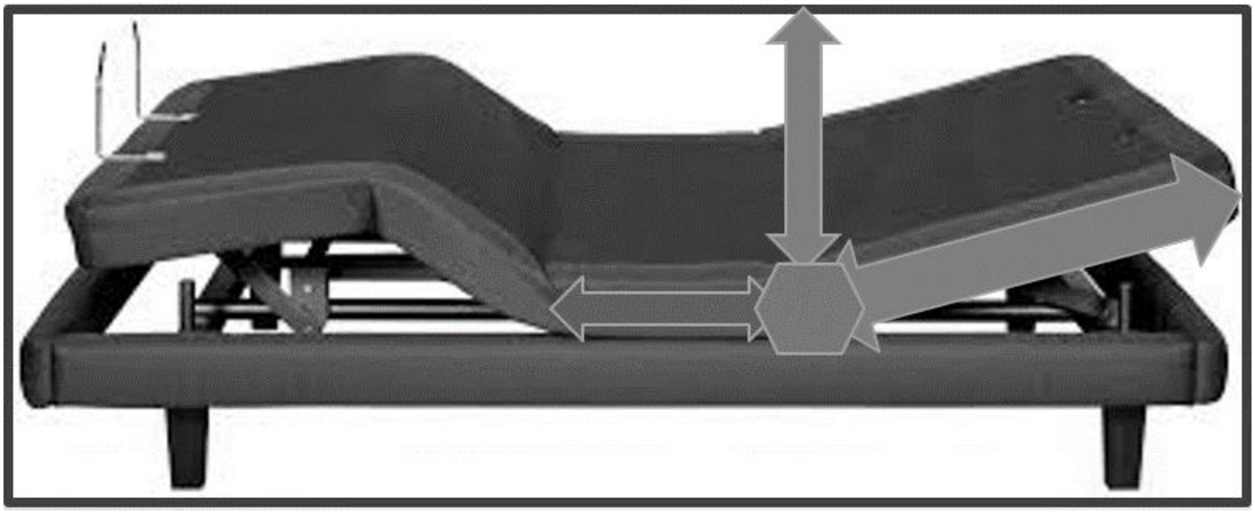


Figure 8 Parts of Adjustable bed

If we consider the above figure. We can see an adjustable bed where the red bad on the right with arrow heads show the upper body whereas the blue bar with the arrow head refers to the pelvic region and the light blue octagon between them depicts the fulcrum or the pivot point between the upper body and pelvic region located at L-1 (Lumber 1). As we can see the upper body is a slab which is hinged are the fulcrum. When the motion occurs in the counterclockwise direction for achieving an upright position. The upper body will reach the other red bar with arrow heads which is at right angle to the blue bar. This would be an upright position for the body.

This is the problem that we face pertaining to lack of mimicking of biomechanics of spine. The spinal column is not just a bar rather a series of joints which create dynamically changing curvatures that keep on changing as motion occurs [19]. Clearly adjustable beds do not provide us with capabilities of following biomechanics of spine while achieving and upright position or vice versa.

As mentioned earlier, we wanted to come up with a solution which provides such motion capabilities while keeping in mind the biomechanics of spine and respecting them during motion/posture achievement.

It is essential to understand the key concepts of kinesiology of spine in order to do so. So in the next section we are going to look at Spine and its kinesiology to get a better grip of the inherent biomechanics of spine so that we can incorporate that into our device [20].

Before that kindly refer to the below figures which pertain to the positioning and profiling that can be achieved with adjustable beds. They also comment on the extreme positions which can be achieved with such beds.

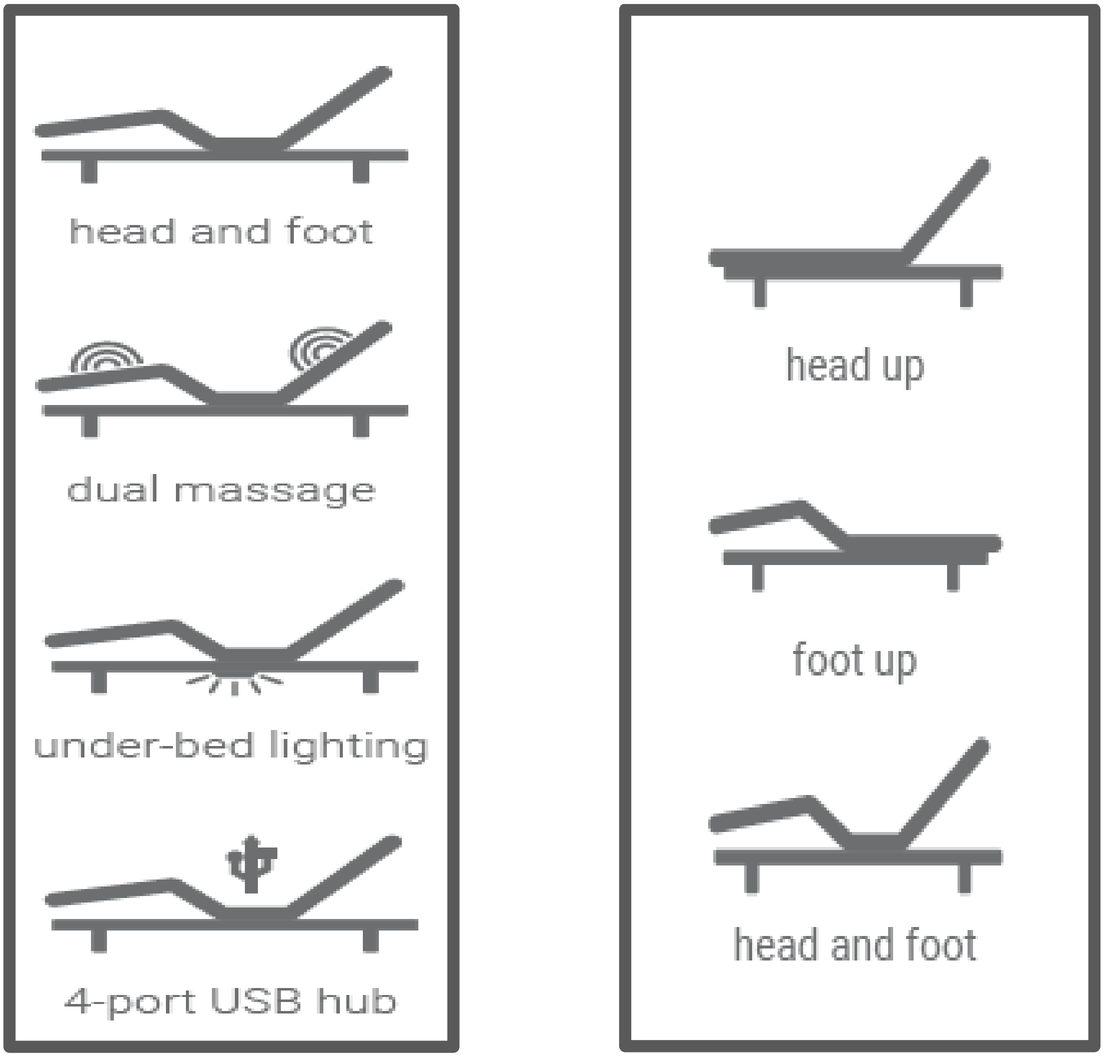


Figure 9 (a) Advanced adjustable bed (b) Simple adjustable bed

2.4 Kinesiology of Spine

2.4.1 Spinal Column Structure

Spinal column structure provides for the basic base of support for the human body. It acts as the linkage between the upper and lower extremities. It also provides protection for the spinal cord which is the pathway for communication between brain and the muscle. Apart from this, it also gives us stability and mobility depending on the type and degree of desired motion. Mobility is referred to as the amount and direct of motion which a segment of spinal column can achieve. Stability is the ability of the spinal column to respond to multiple forces which are place on it while maintaining its structural integrity.

2.4.2 Regions of Spinal Column

There are 5 distinct regions of the spinal column. The cervical region or segment refers to the neck region. It consists of 7 vertebras. Thoracic region is the chest region. There are 12 vertebras in this segment. The abdominal region of the body is provided support by the lumber region of the spine. It consists of 5 vertebras.

The pelvic region composes of amalgam of fused Sacral bones and the coccygeal bone. In total, there are 33 bones and 23 disks between them.

2.4.3 Spinal Curvatures

Before the birth of human beings, the spinal column structure is a C-shaped structure which as the human grows firstly straightens and then converts into distinct curvatures which at adulthood leaves us with 4 distinct curvatures.

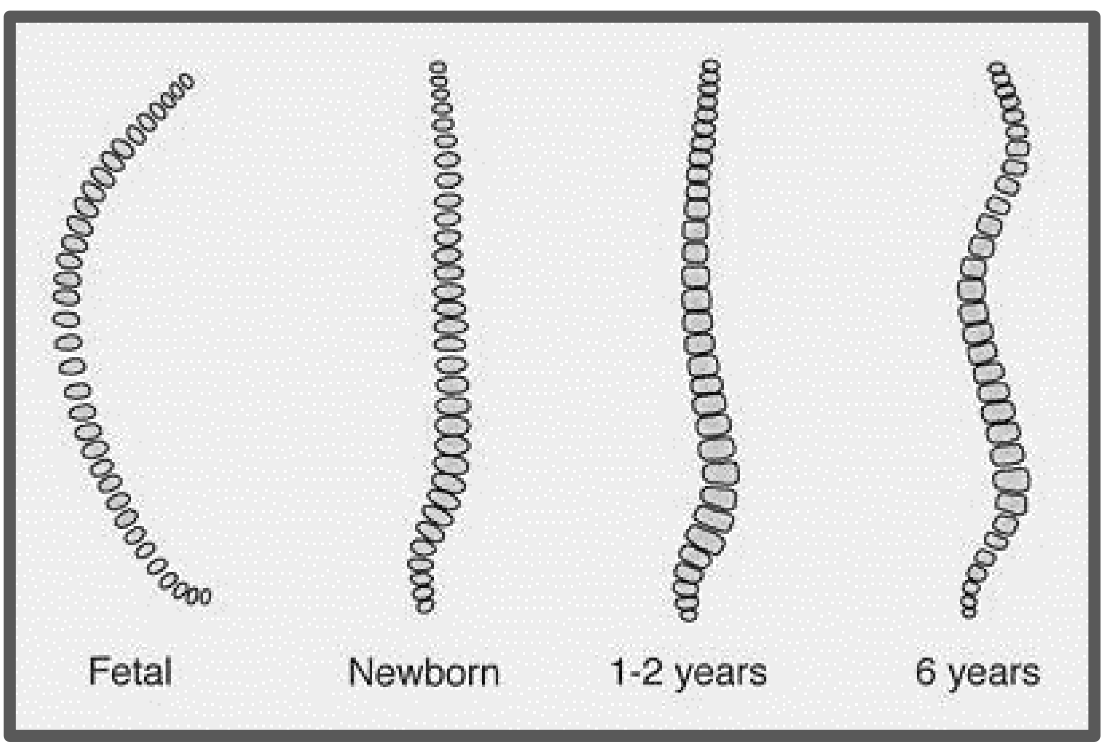


Figure 10 Spinal Curvature

Out of these 4 distinct curvatures, one is fixed as it consists of fused bones whereas the rest of the three have the capabilities of dynamically changing curvatures. It is since all these dynamically changing curvatures are made by segments of spinal column where there are joints and disks between adjacent vertebras providing for motion.

The three dynamically changing curvatures are named as

- Cervical Lordosis
- Thoracic Kyphosis
- Lumber Lordosis

Refer to the figure below to understand the positioning and locations of these curvatures.

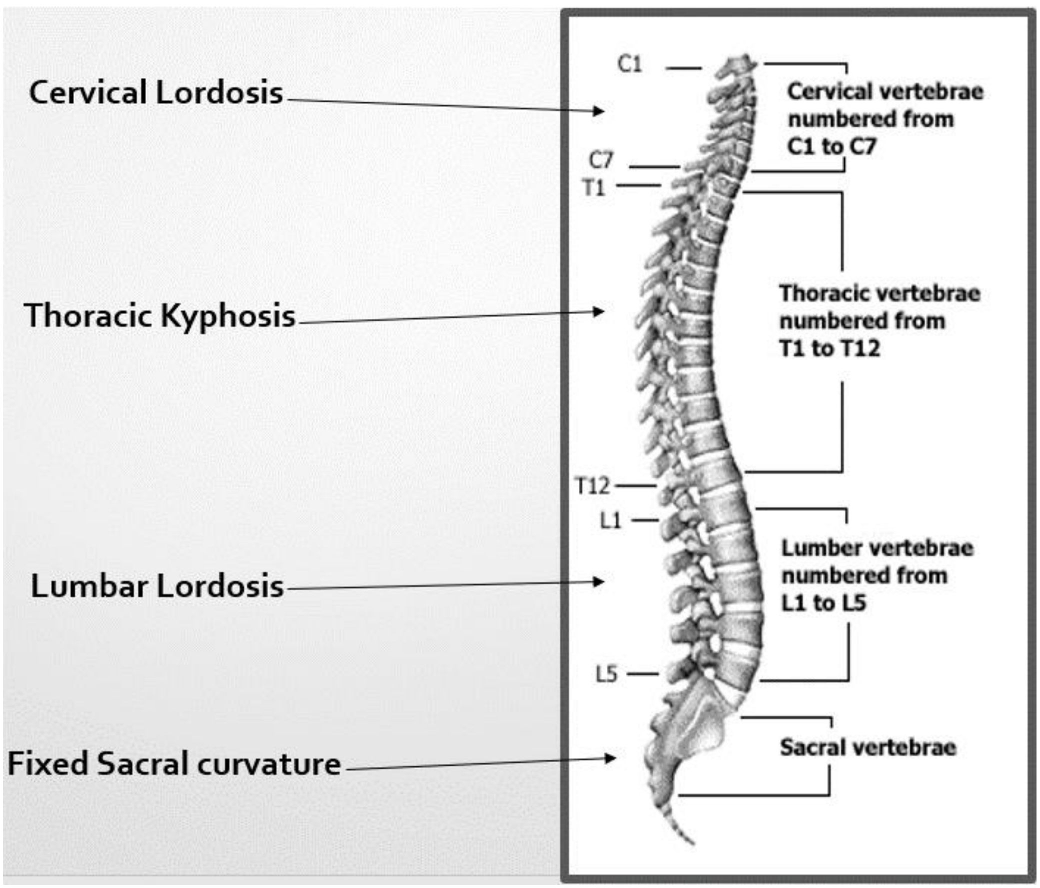


Figure 11 Positioning and Location of Spinal Curvatures

2.4.4 Cervical Region

The cervical region of the spine consists of 7 vertebras. The top most vertebra of this region makes the atlanto-occipital joint which is the connection between the human skull at occipital lobe and human spine. The vertebras of this region are smallest in comparison to the rest of the spinal vertebras. Due to small size this region provides most mobility and in turn least stability. The nature of small vertebras gives this region very low load bearing capabilities.

2.4.5 Thoracic region

There are 12 vertebras in this segment of the spine. On a comparative scale the vertebras of this region are larger than the vertebras in the cervical region. Due to larger vertebras the load bearing capabilities of this section is higher and mobility in this region is low. It has the least mobility among all the regions of the spine which is partly due the articulation of the ribs. Articulation of the ribs greatly increases the load distribution hence providing for more stability in this region as well.

2.4.6 Lumber region

The lumber region has 5 big vertebras [23]. It has the largest vertebras amongst all the sections of the spine. Due to large size of the vertebras this region provides more stability compared to mobility. Lumber spine has the highest load bearing capabilities [24].

2.4.7 Spinal movement

Spinal movement is due to the integration and combination of intervertebral and facet joints which are controlled by the spinal musculature located at human back and upper torso [22]. As a result, human back can achieve three types of rotation which are as follow

- Rotation along Antero-posterior axis
- Rotation along Medio-lateral axis
- Rotation along transverse axis

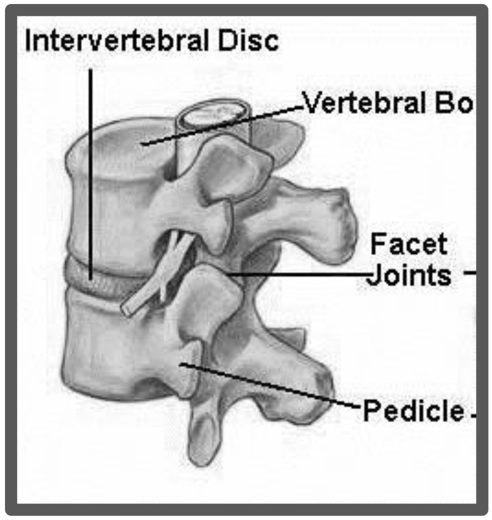


Figure 12 Parts of Spinal Joint

For the scope of this research we took in to account the first two rotations [25]. As we know the task for this research was to come up with a device which could allow for bed ridden patients to sit from a lying posture and vice versa while keeping in mind the biomechanics of spine and comfortability of the patient as well.

So essentially, we wanted to achieve rotation along the medio-lateral axis through the actuation of the device while providing for flexibility in the anteroposterior axis [21]. The following figure would better depict the context of the above statement.

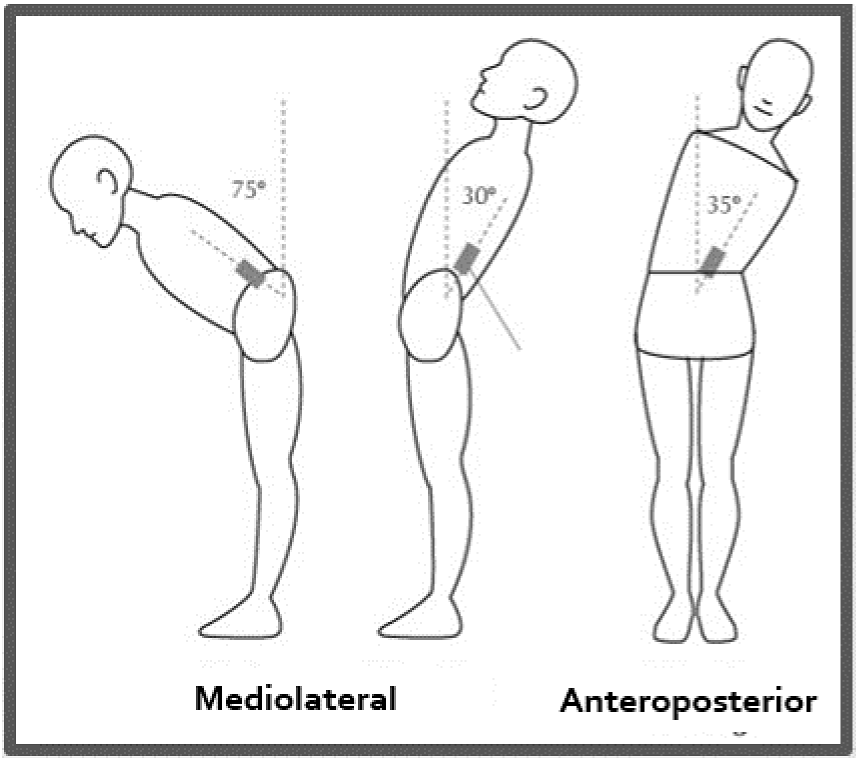


Figure 13 Rotations of Spine

This brings end to the second chapter of this thesis. It encompasses all the literature which was referred to get the key concepts which were essential in the conception of the design. Now we are going to move towards the third chapter which pertains to the development of the device.

CHAPTER 3 Device Development

3.1 Plate and Hinge Mechanism

As mentioned at the end of the previous chapter. We were interested in achieving two types of rotation with the mechanical assembly namely

- Rotation along medio-lateral axis
- Rotation along Antero-posterior axis

To achieve such motions, we came up with a passive assembly comprising of a series of plates and hinge mechanism [26]. As mentioned earlier spinal column has different sections which have different mechanical limitations to mimic these mechanical limitations three distinct types of plates were used for the passive assembly. Each representing a specific region of the spine. The three types of plates fabricated were as follow

- Cervical plate
- Thoracic plate
- Lumber plate

Each plate typed differed from others based on dimensions/size. All the plates were made of Aluminum and covered with padding and latherite material. Please find below pictorial depiction of the plates.

Figure 14 Cervical Plate

Figure 15 Thoracic Plate

Figure 16 Lumber Plate

In between two adjacent plates a hinge joint was kept which allows each plate with two degrees of freedom (rotation along medio-lateral axis and rotation along antero-posterior axis) similar to as provided by the human spine [27].

Find below the figure showing the hinge joint

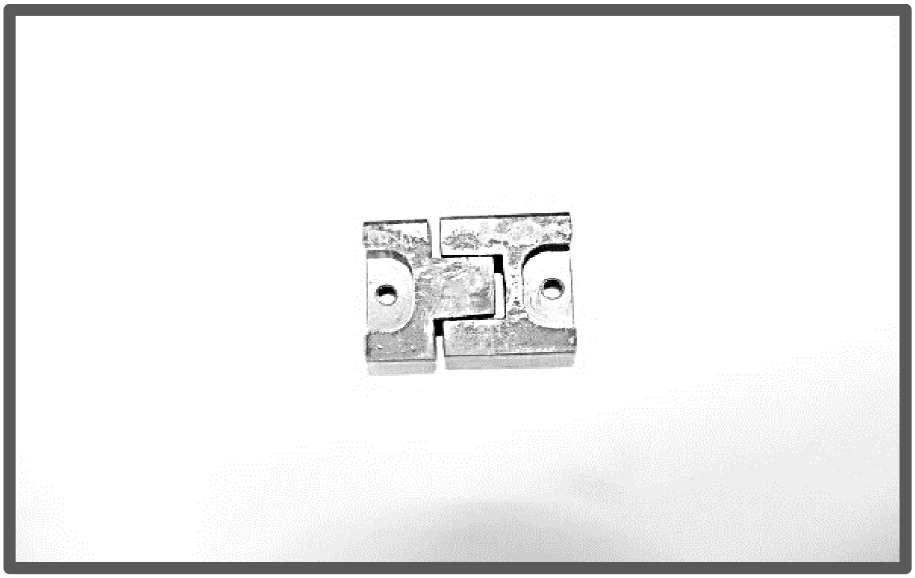


Figure 17 Mechanical Hinge Joint

As the above figure suggest the hinge joint comprised of two parts namely

- Hinge Part 1
- Hinge Part 2

Both parts were joined together using S.S wire of dia 4mm. Each hinge part had a U channel with a hole in it where the plates were screwed. The following figure shows the assembled passive plate and hinge mechanism

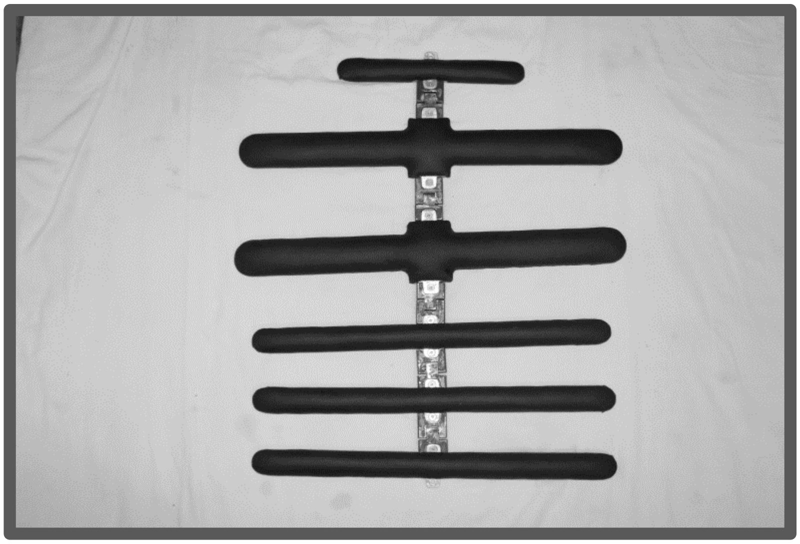


Figure 18 Passive Plate and hinge assembly

3.1.1 Rotation in the antero-posterior axis

To achieve the flexibility in the antero-posterior axis the U channels on the hinge joints were kept marginally bigger than the male U channels of the plate. This allowed for restricted rotation in the antero-posterior axis in both clockwise and anti-clockwise direction as shown by the following figures

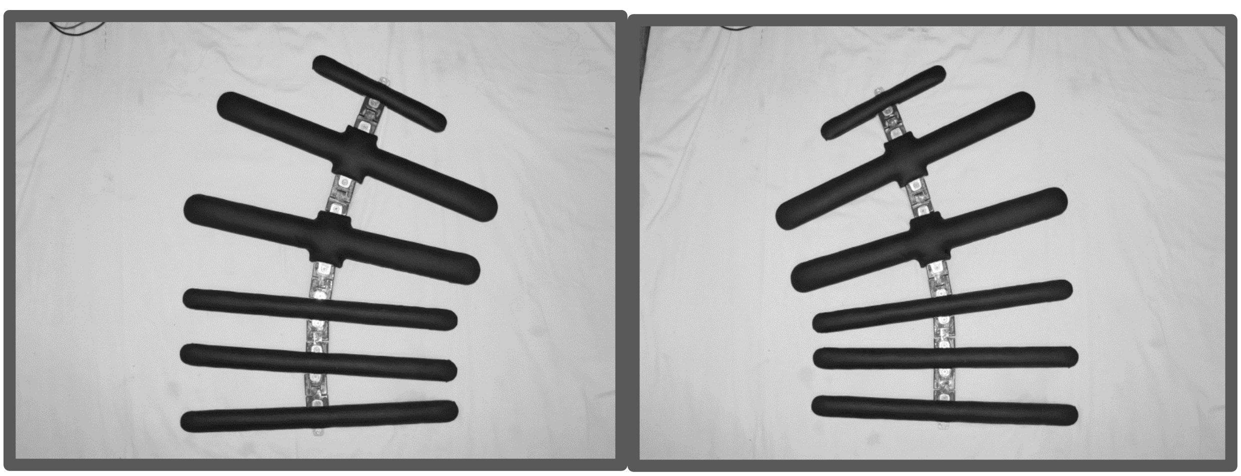


Figure 19 (a) Clockwise direction (b) Anti-clockwise direction

As the above figures clearly suggest that the passive assembly provides for the desired flexibility in the antero-posterior axis [27]. Biomechanics of the spine are being incorporated in the design.

Next, we are going to look into the rotation capabilities of the passive assembly in the mediolateral axis.

3.1.2 Rotation in the Medio-lateral axis

To achieve rotation in the mediolateral axis while respecting the biomechanics of the spine. It was essential to understand the spinal curvatures and normality pertaining to them. For this would help us in setting the mechanical constraints on the extreme limits of the mechanism.

3.1.2.1 Normal Spinal Curvatures

As we have already discussed there are 4 distinct curvatures which a human spinal column makes. Out of these four only three have the capability to change dynamically as a motion is achieved. Even though these curvatures change their positioning with respect to each other. They still have some limitations and constraints that are being followed. We also know that the cervical region of the spine provides most mobility due to their small sized vertebras.

According to literature review we were able to pick the following ranges for normality of the spinal curvatures

- normal range for Thoracic kyphosis is between 20°-45°. It is marginally normal btw 45°-55°. Any angle below 20° is an abnormality and is referred to as hypo-kyphosis, similarly if the angle is above 55° it is termed as hyper-kyphosis which is commonly known as ‘round back’.
- Normal levels for lumbar curvature, ‘Lumber lordosis’ is between 20°-55°. Reduced angles are referred to as a condition of hypo-lordosis, similarly increased angle beyond the asserted range above lead to a human condition known as hyper-lordosis, commonly known as Flat back.

The following figures comment on the normality pertaining to the spinal curvatures.

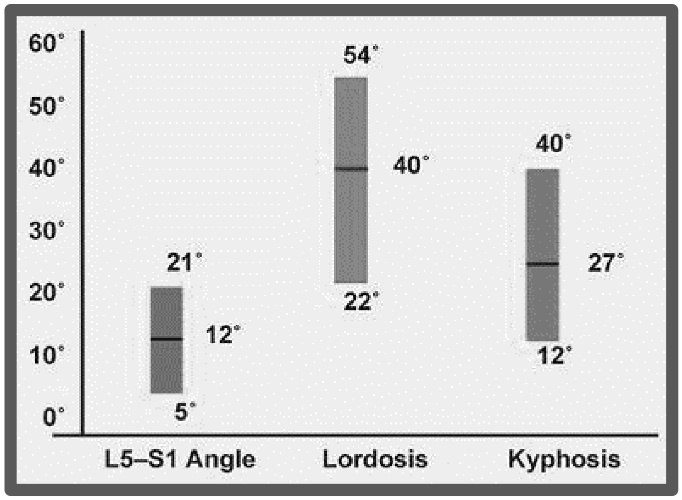


Figure 20 Graph showing Normal range of curvatures of spine

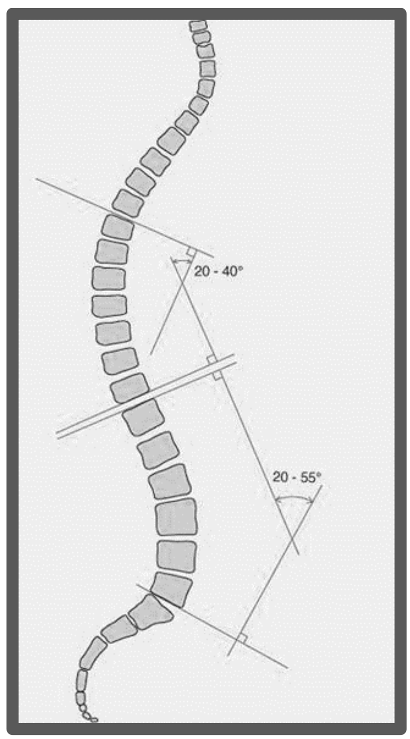


Figure 21Extreme angles of normality

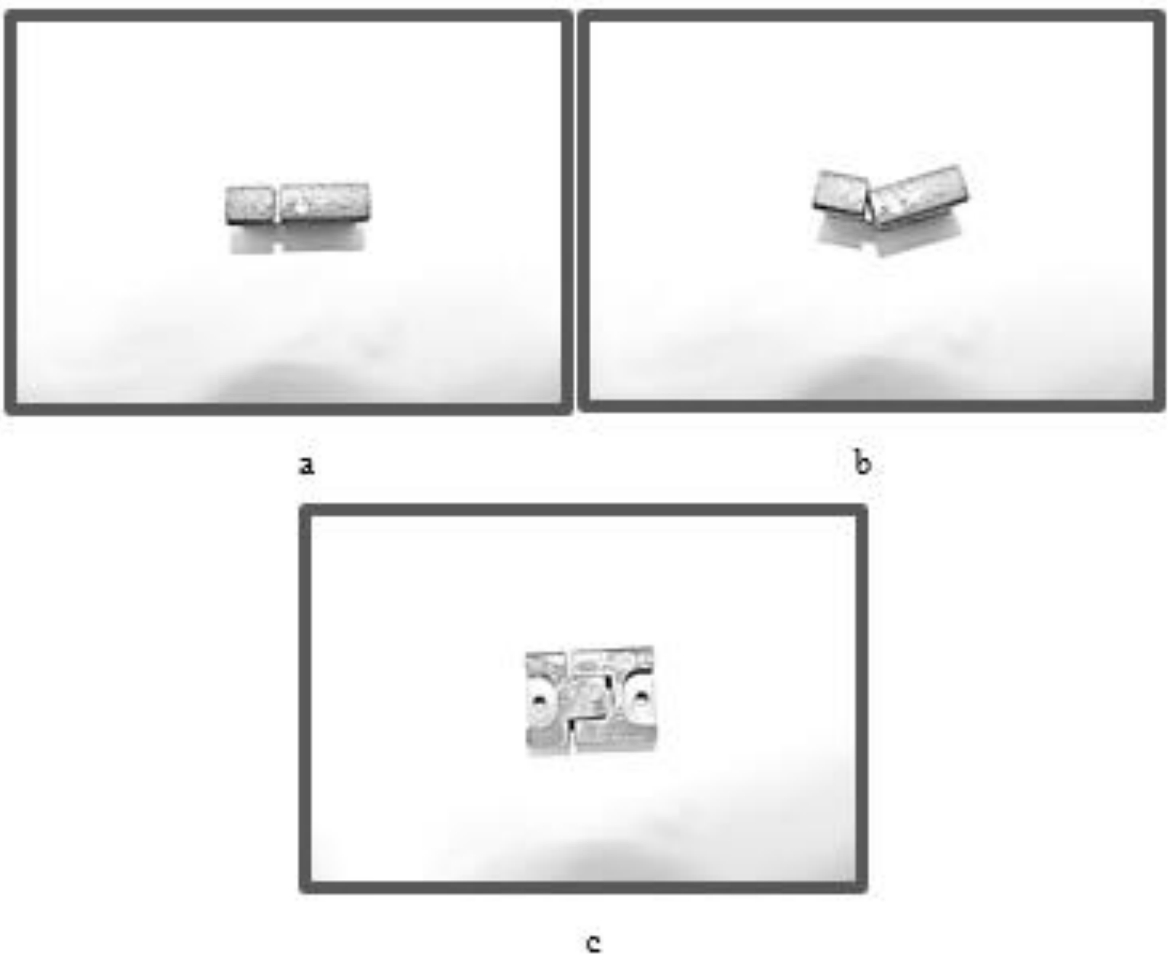
From the above figures we were able to take out the extreme angles which can be achieved by the spine. This data was essential for us to understand the inherent biomechanics of spine in the medio-lateral axis as our goal was to mimic the spinal kinesiology in this axis to allow for patients to sit up right from a lying posture while respecting the normality of spinal curvature in this axis.

3.1.2.2 Mechanism features

As mentioned earlier we used a passive assembly of series of plates and hinge mechanism to provide for the desired actuation. There were two rotations that we wanted to achieve. Rotation in the antero-posterior axis and rotation along the medio-lateral axis [28]. We have already commented on the antero-posterior axis. Now, we are going to discuss how this passive assembly achieves the rotation in the medio-lateral axis while respecting the normality of spinal curvatures in doing so.

To achieve this, we used hinge joints with variable limits so that we could provide rotation at each joint respecting the degree of rotation required on each joint in accordance to the normality of curvatures at that region of spine.

You can find below the pictures of hinge joint.



a b

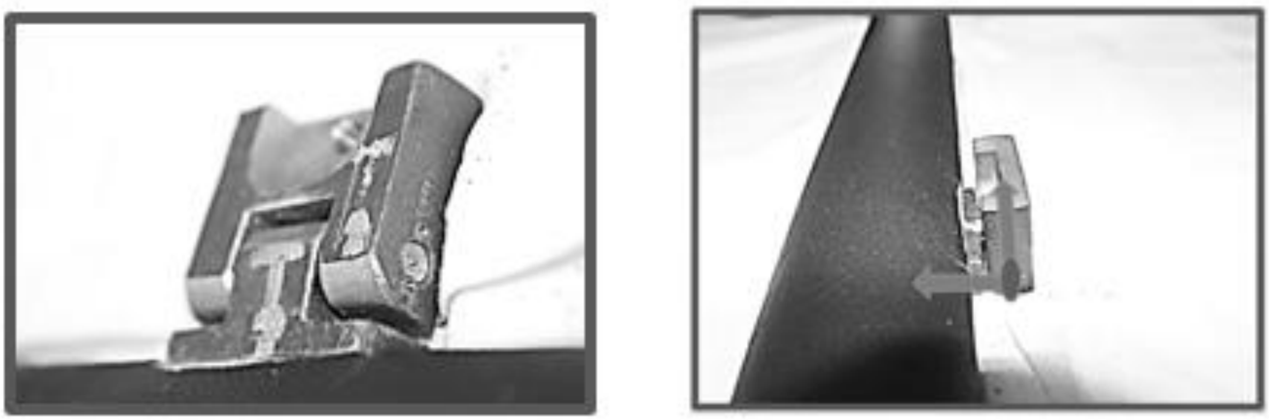
c

Figure 22(a)Side view: no rotation (b) Side view: rotation (c) Top View

Now, we are going to investigate the details of the variable hinge joints used at different regions of spine, starting from the sacrum region up to the parietal lob of the skull.

3.1.2.2.1 Hinge at Sacrum region

It is the first hinge joint. It is located are the meeting point of sacrum region with the lumber region. This joint allows for maximum 90 deg of rotation in the clockwise direction. Refer to the following figures to get a better idea.



a b

Figure 23 (a) 0 degree (b) 90 degree

3.1.2.2.2 Hinge between L1 and L2

It is the second hinge joint. It is located between the Lumber1 to Lumber3 vertebras within the lumber region. This joint allows for maximum 15 deg of rotation in the clockwise direction. Refer to the following figure to get a better idea.

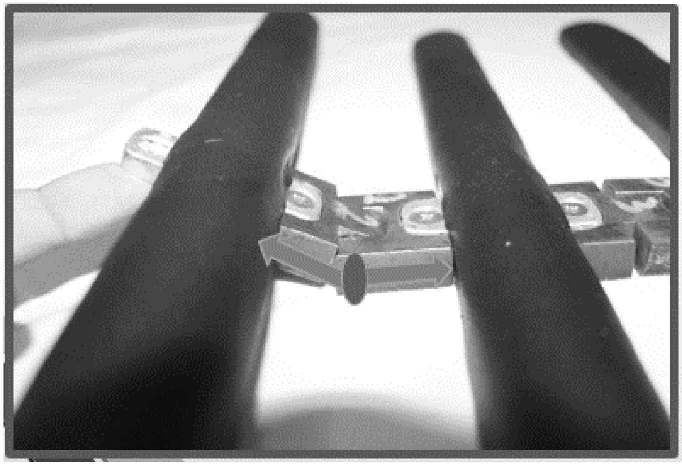


Figure 24 15 degree

3.1.2.2.3 Hinge between L2 and L3

It is the third hinge joint. It is located between the Lumber3 to Lumber5 vertebras within the lumber region. This joint allows for maximum 50 degree of rotation in the clockwise direction. Such high value for angle projection is due to the face that the max angle of deviation in angle can only be achieved at this region of the spine. Refer to the following figure to get a better idea of angle projection at this joint.

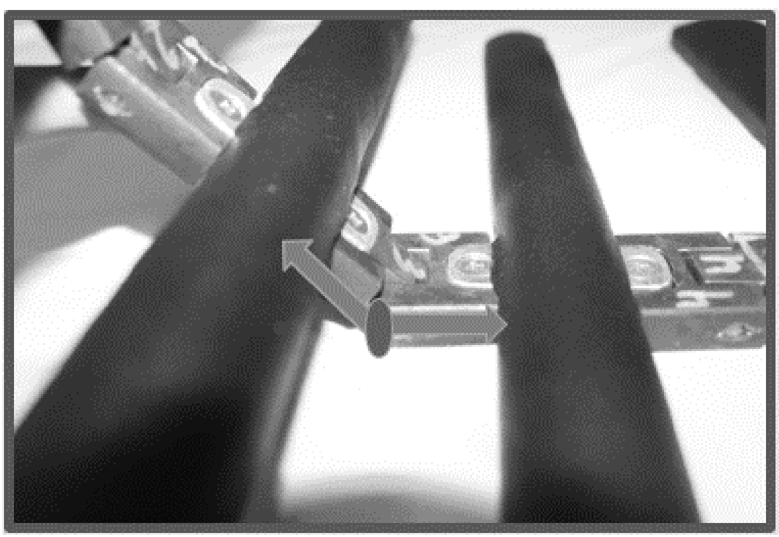


Figure 25 50 degree

3.1.2.2.4 Hinge at between L3 and T1

It is the fourth hinge joint. It is positioned between the Lumber5 to Thoracic1 vertebras. This joint is the linkage between lumber and thoracic region. This joint allows for maximum 10 degree of rotation in the clockwise direction. Refer to the following figure to get a better idea.

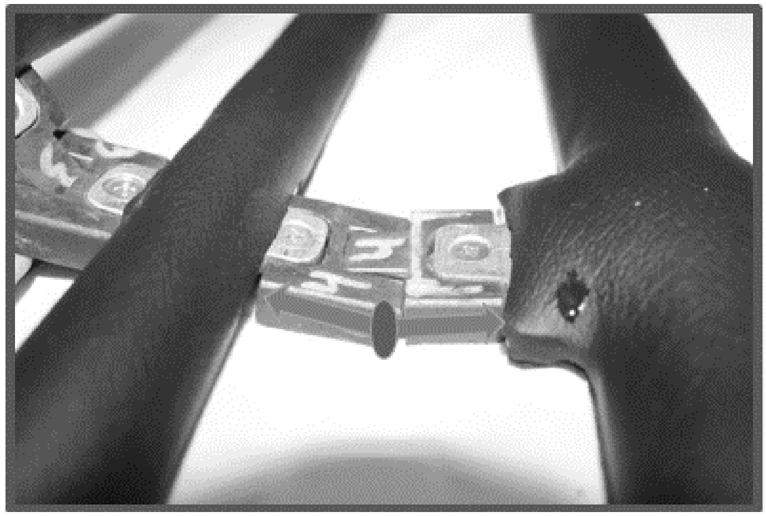


Figure 26 10 degree

3.1.2.2.5 Hinge between T1 and T2

It is the Fifth hinge joint. It is the only joint located between the Thoracic1 to Thoracic12 vertebras within the thoracic region. As we know due to the articulation of ribs, the mobility is least in this region. So, to mimic such restrictions this joint was fabricated in a manner to allow for only a maximum 5 degree of rotation in the clockwise direction. Refer to the following figure to get a better idea.

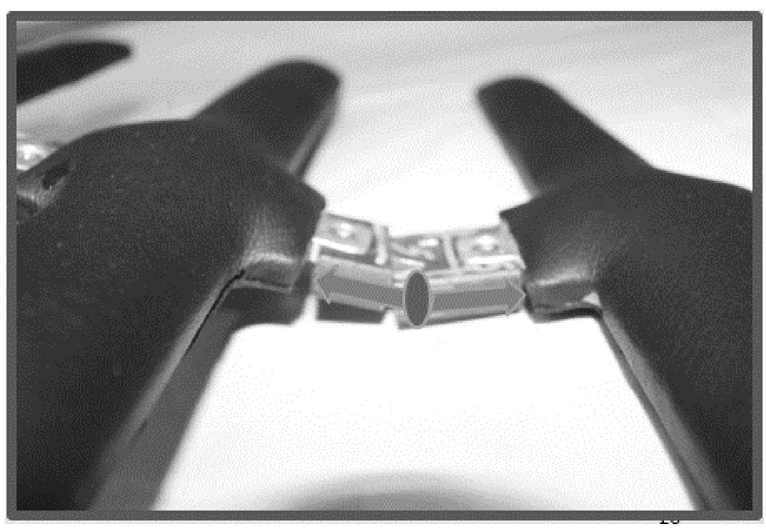


Figure 27 5 degree

3.1.2.2.6 Hinge between T2 and C1

It is the sixth hinge joint. It is positioned between the Thoracic12 to cervical3 vertebras. This joint is the linkage between Thoracic and cervical region. This joint allows for maximum 10 deg of rotation in the clockwise direction. Refer to the following figure to get a better idea.

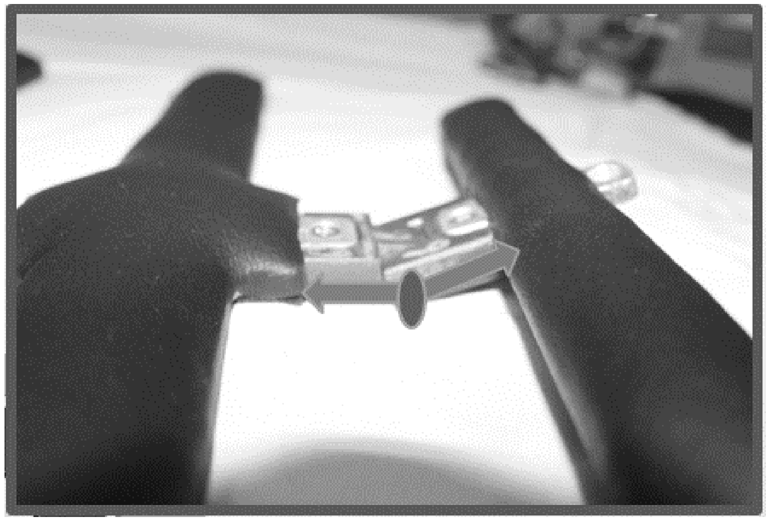


Figure 28 10 degree

3.1.2.2.7 Hinge between C1 and C2

It is the seventh hinge joint. It is located between the Cervical3 to Cervical7 vertebras within the cervical region. This joint allows for maximum 10 deg of rotation in the clockwise direction. Refer to the following figure to get a better idea.

Figure 29 10 degree

3.1.2.2.8 Head Rest

The head rest consisted to two plates fixed with each other using a rigid joint. One of the plates was C2 and the other was Top plate. Due to the rigid joint there was no degree of freedom between these two plates (no provision for rotation as with other joints). Find below the pictorial depiction of the head rest.

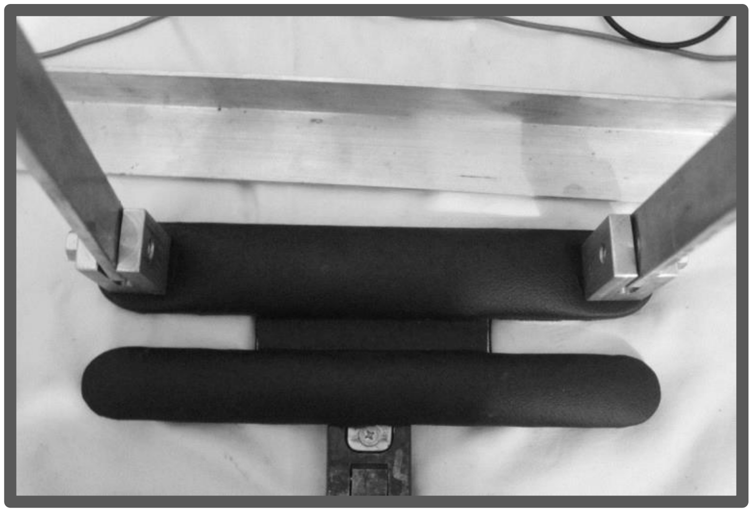


Figure 30 Head Rest

3.2 Design parameters

Now that the details of the passive assembly have been discussed. It leads us to the conception and development of the active part of the device. The active part of any device is the one which provides for the necessary actuation required for the working of the device [29].

Before we move ahead with the conception and development of the active part. We need to set a few design parameters. These design parameters are essential considerations for giving us the specifications required by the device for successful working.

Major design parameters which were considered for the successful working of the device have been quoted below

- Maximum load to be lifted
- Required torque

3.2.1 Weight considerations

Weight is one of the most essential consideration for designing of a device, the desired action of which is to bear and lift it. So, we needed to set the maximum limit of weight that the device would be able to lift.

For this design parameter we needed to consult literature and accordingly set the maximum limit of the weight that the device would be able to lift. Firstly, we need to realize that the human body can be segmented on a macro level in the following constituents

- Upper body (includes the upper limb)
- Lower body

According to human body studies we were able to conclude that the following weight ratios

- The upper body makes up to 55% to 60% of the whole weight of the human body
- The lower body ranges from 40% to 45% weight of the whole body

The next step was to find how much does a human body weight. To find that out we had to consult the literature for ideal weight of human body for different heights. For this allowed us to get a better estimate of the required weight to be lifted by the mechanism.

The following table comments on the ranges of ideal range of weight for men and woman based on their height.

MALE		FEMALE	
Height	Ideal Weight	Height	Ideal Weight
4' 6"	28 - 35 Kg	4' 6"	28 - 35 Kg
4' 7"	30 - 39 Kg.	4' 7"	30 - 37 Kg.
4' 8"	33 - 40 Kg.	4' 8"	32 - 40 Kg.
4' 9"	35 - 44 Kg.	4' 9"	35 - 42 Kg.
4' 10"	38 - 46 Kg.	4' 10"	36 - 45 Kg.
4' 11"	40 - 50 Kg.	4' 11"	39 - 47 Kg.
5' 0"	43 - 53 Kg.	5' 0"	40 - 50 Kg
5' 1"	45 - 55 Kg.	5' 1"	43 - 52 Kg.
5' 2"	48 - 59 Kg.	5' 2"	45 - 55 Kg.
5' 3"	50 - 61 Kg.	5' 3"	47 - 57 Kg.
5' 4"	53 - 65 Kg.	5' 4"	49 - 60 Kg.
5' 5"	55 - 68 Kg.	5' 5"	51 - 62 Kg.
5' 6"	58 - 70 Kg.	5' 6"	53 - 65 Kg.
5' 7"	60 - 74 Kg.	5' 7"	55 - 67 Kg.
5' 8"	63 - 76 Kg.	5' 8"	57 - 70 Kg.
5' 9"	65 - 80 Kg.	5' 9"	59 - 72 Kg.
5' 10"	67 - 83 Kg.	5' 10"	61 - 75 Kg.
5' 11"	70 - 85 Kg.	5' 11"	63 - 77 Kg.
6' 0"	72 - 89 Kg.	6' 0"	65 - 80 Kg.

Figure 31 Ideal weight to height

Consulting the above-mentioned table, we were able to estimate the weight considerations for the device actuation. The device was designed for an ideal height of 5'11''. As the table mentions, the range of ideal weight for such human is 70-85kg for males and 63-77kg for females respectively. Getting an average of these ranges we were able to come up with an average weight of 75kg. This was taken as the maximum limit of weight to be lifted.

Consulting the literature furthermore, we were able to estimate the distance between lumber1 vertebra to the top of parietal lobe to be a range between 65 to 70 centimeters for a human body of 5'11'' height. This distance was an important parameter to consider as it was a prerequisite in defining the torque requirements.

3.2.2 Torque calculations

As already discussed, for the working of device, an active component was required for the actuation of the mechanism which would provide the necessary force required for achieving the postural change. It is also in our knowledge that the degrees of freedom that the device provided for was in the form of rotation. This brings in the concept of torque which is the required force that can rotate an object about an axis or pivot. Like force which brings about linear acceleration in an abject, Torque causes an object to acquire angular acceleration. It is noteworthy that Torque is a vector quantity meaning both direction and magnitude affect its resultant outcome.

It is governed by the following formula

$T = r \times F$

Where 'T' is the resultant torque. 'r' is the moment arm which is the perpendicular distance between the pivot point/axis around which the body rotates and the point where the force 'F' is being applied on the body.

Now considering that we set the max load of the human body to be lifted, at a limit of 75kg and taking into account the fact that 55% to 60% of the body weight comprises of the upper body. It can be clearly deduced that sitting upright for a person requires force propagation/initiation in the upper body only. So, we took 45kg to be the load which needed to be lifted. As it is 60% of 75kg. The reason for taking 60% rather than 55% was to better the torque estimate in the upper limit. If we design our system according to 60%, it will easily be able to work successfully for 55% as well. Refer to the below figure to better grasp the concept applied.

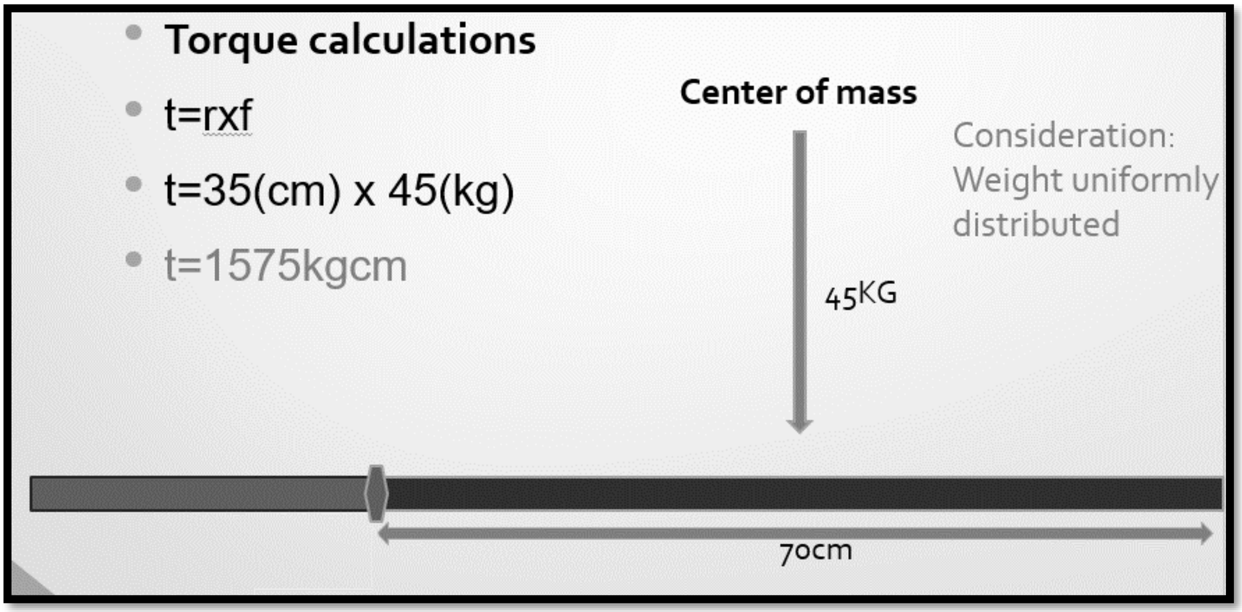


Figure 32 Torque calculations

if you look at the above figure you can see two bars are shown. The grey bar represents the lower body and the blue bar shows the upper body while the light blue octagon shows the pivot or fulcrum at the pelvic girdle. According to the above consideration we know that the upper body weighs 45kg. This blue bar rotates at the fulcrum, in anticlockwise direction to achieve an upright posture. The length of the blue bar is taken as 70centimetes. Here we have taken an assumption that the weight on the bar showing the upper body is uniformly distributed. In a bar with uniform weight application of load is at the midpoint of the bar. This midpoint represents the center of mass of the bar (COM). Center of mass is the point on the body where the load is applied. Taking in to account the length of the bar representing upper body we know the center of mass is at 35 centimeters. This distance is the moment arm 'r' in the torque formula while 45kg is the force being applied. Putting in these values in the formula of torque we get the required torque which is 1575kgcm.

So, to bring about rotation in a straight bar this much torque is required. Even though the weight of human body is not uniformly distributed. This straight bar assumption gives us an estimate of the torque that will be required for the desired postural changes [30].

As we know ‘r’ and ‘F’ are directly proportional to Torque. Decrease in the moment arm or in the applied force are going to lower torque requirements. If we shift the fulcrum from pelvic girdle towards the parietal lobe. Two things are going to happen.

- Moment arm is going to decrease lowering torque requirements
- Force or load is going to decrease (as we have considered the uniform weight distribution) lowering torque requirements

Now, we are going to bring in such shifts and see how it affects torque requirements for our system. We are going to bring a change of 10cm each time and calculate the torque incorporating those changes. As mentioned earlier, we have considered the upper body to be a bar with uniformly distributed weight. With each shift of 10cm in the fulcrum the load on the rotating segment is going to lower by 6.42kg.

Refer to the following figures pertaining to such shifts and their effects on the required torque

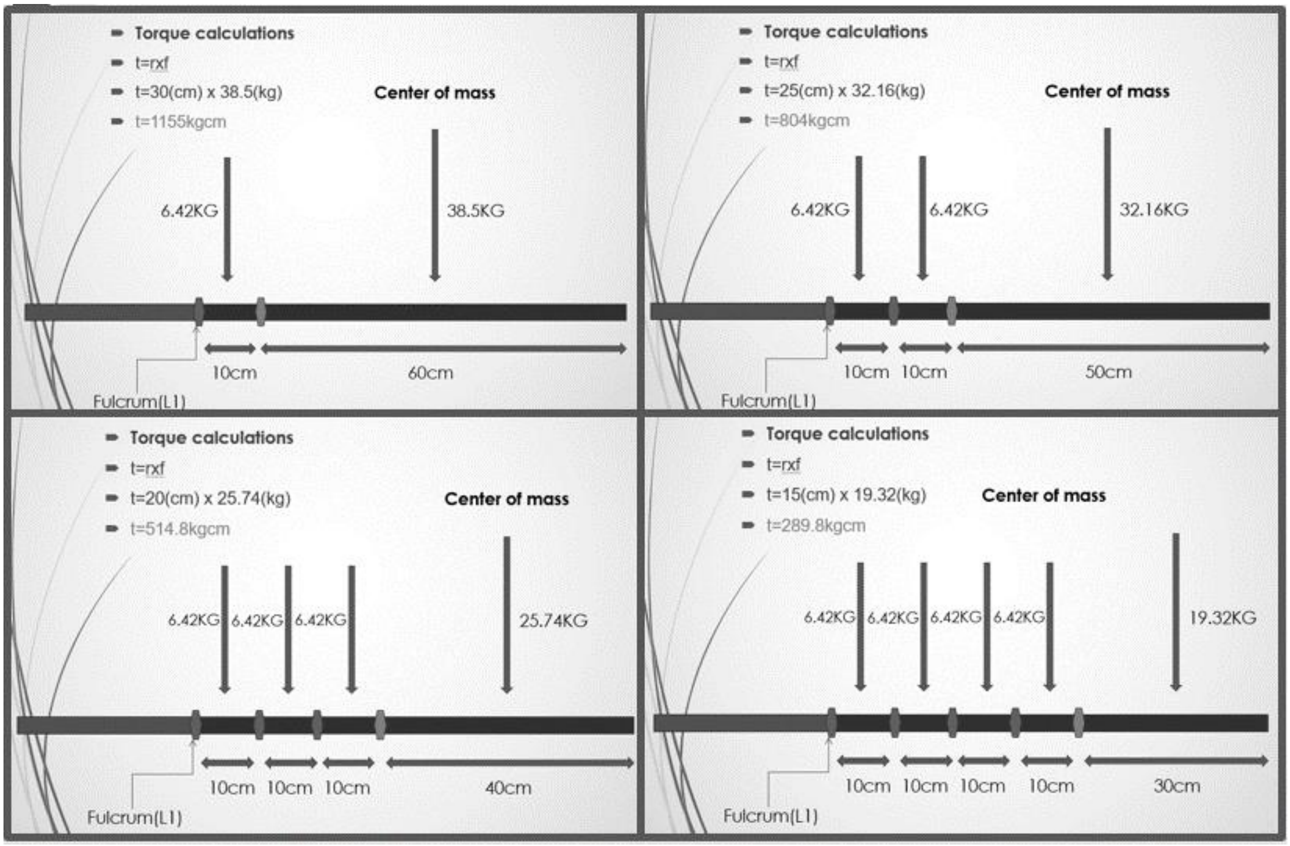


Figure 33 Effects on torque with shift of 10cm in the fulcrum or the pivoting point

As we see from the above figures,

- with only one shift of 10cm the required to torques decreases to 1155kgcm.
- With two shifts of 10 cm the required torque decreases to 804kgcm
- With three shifts of 10cm the required torque decreases to 514.8kgcm
- With four shifts of 10cm the required torque decreases to 289.8kgcm

It is noteworthy that for above calculations we assumed the upper body to be a bar with uniform load acting on it, making the midpoint of the bar the acting Center of Mass.

For the above torque calculations, we have considered the bars to be horizontal, making a projection of 0deg with the axis of rotation. These calculations do not encompass the effects of changes in angle projection.

When we take shift of angles in to account, Torque is governed by the following formula

- T=r x F cos(theta)

Where theta is the angle that the bar is making with the axis point in our case. For the extreme positions to be achieved are sitting upright and lying down. The limits for theta are as follow

- Theta= 0-90deg

The function of COS gives the following estimations

- Cos (90) = 0
- Cos (0) = 1

So, as Cos (0) gives us 1, the highest torque requirements for the device would be when the human body is in the lying posture. It would show a trend of decrease as the body progresses to achieve an upright position (given by Cos (90) = 0).

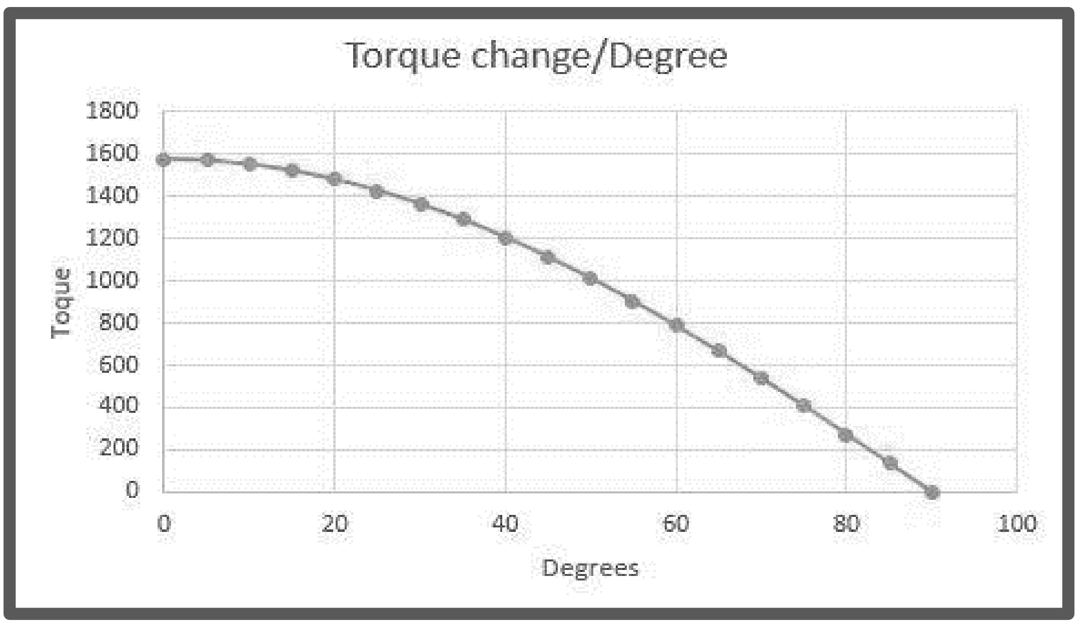


Figure 34 70cm

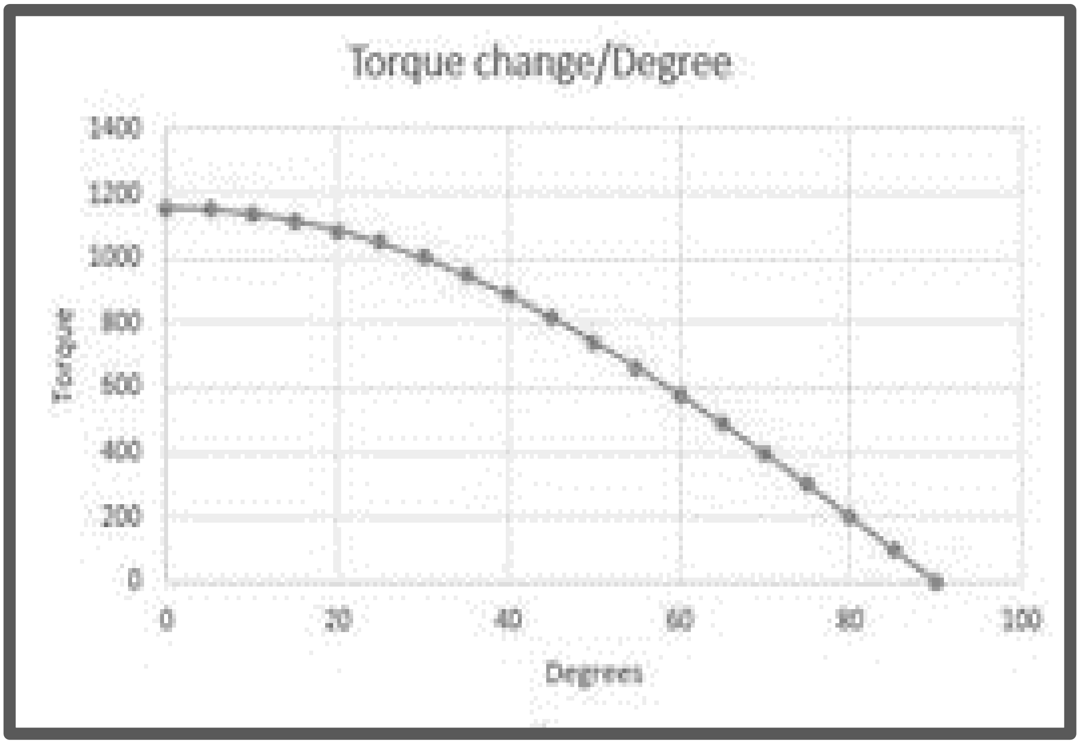


Figure 35 60cm

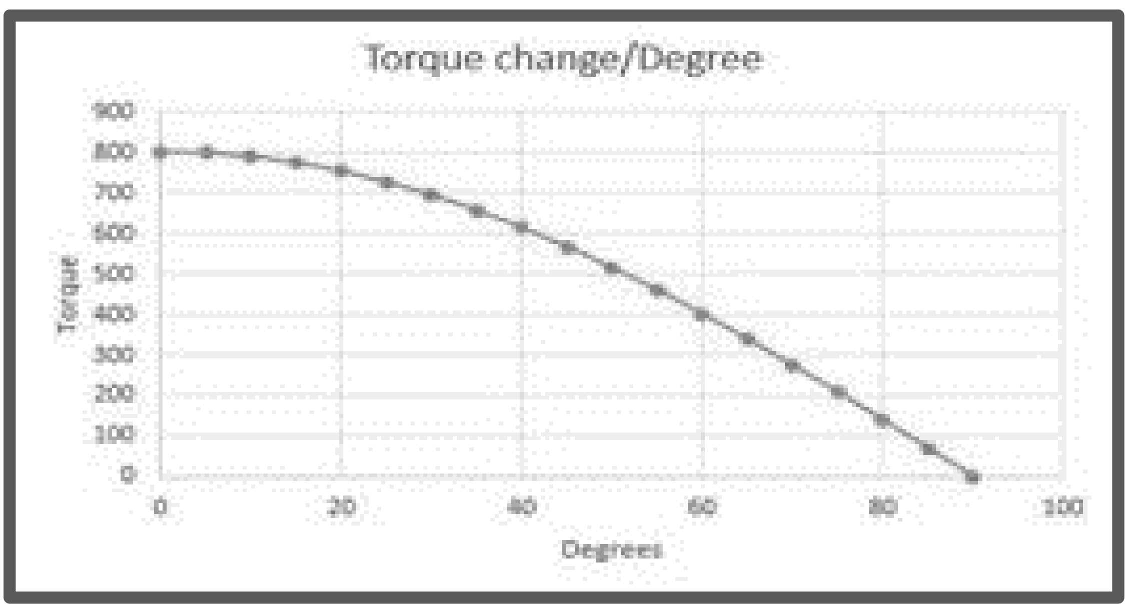


Figure 36 50cm

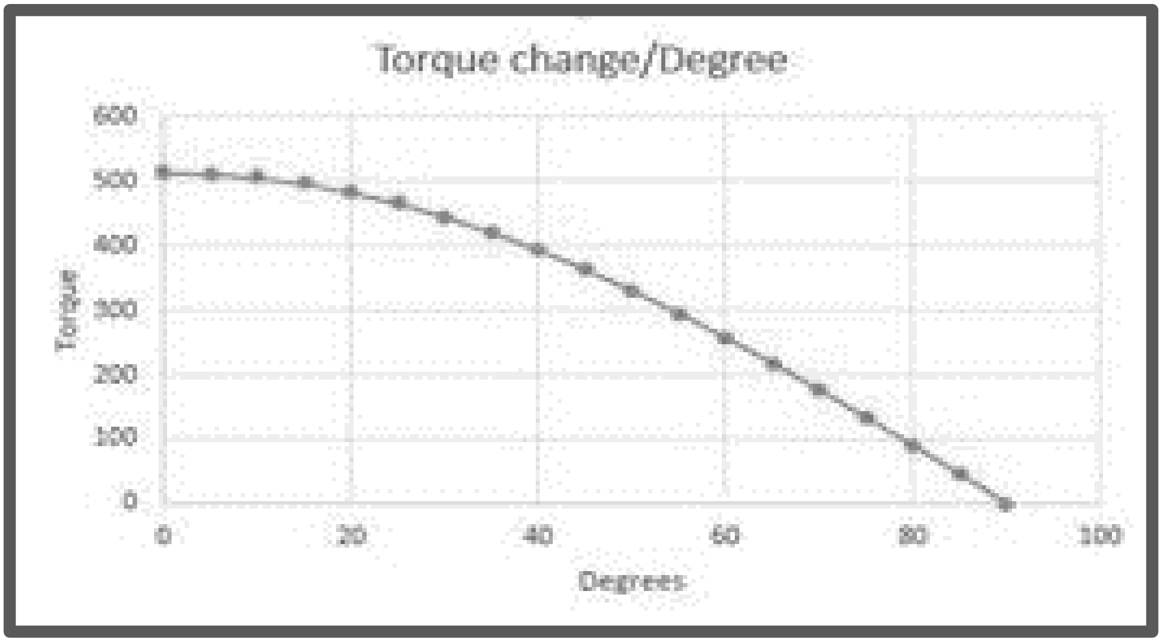


Figure 37 40cm

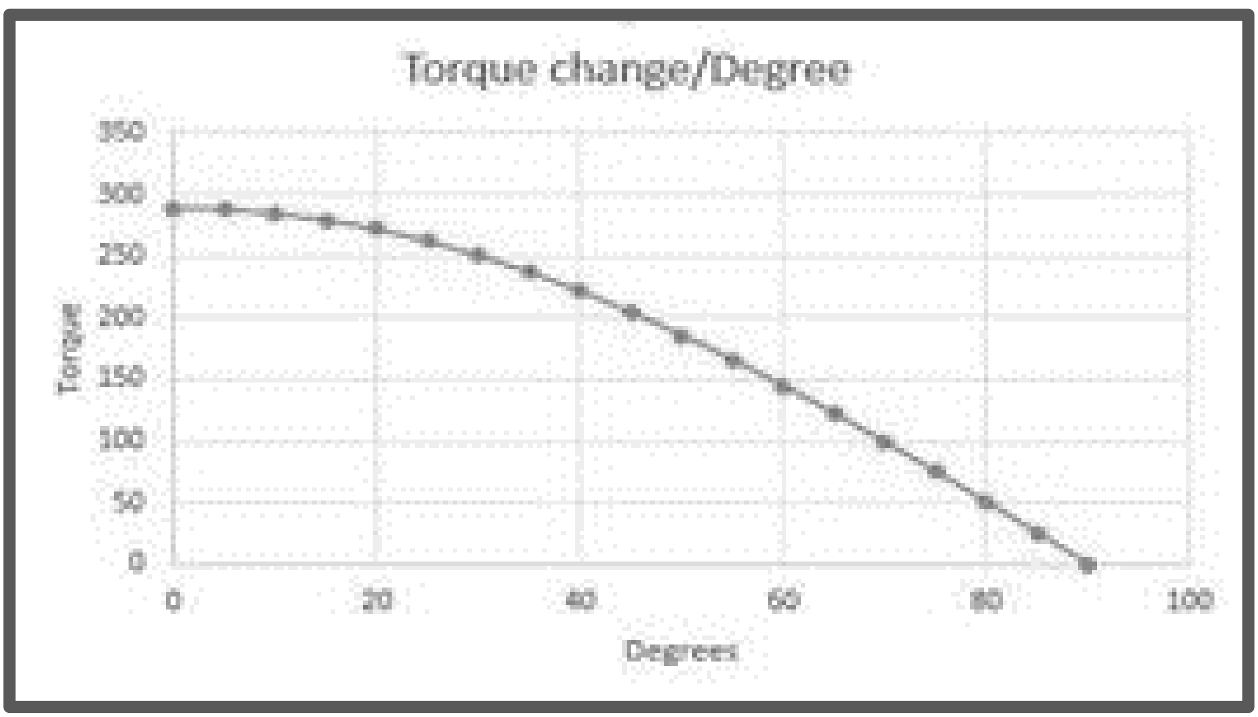


Figure 38 30cm

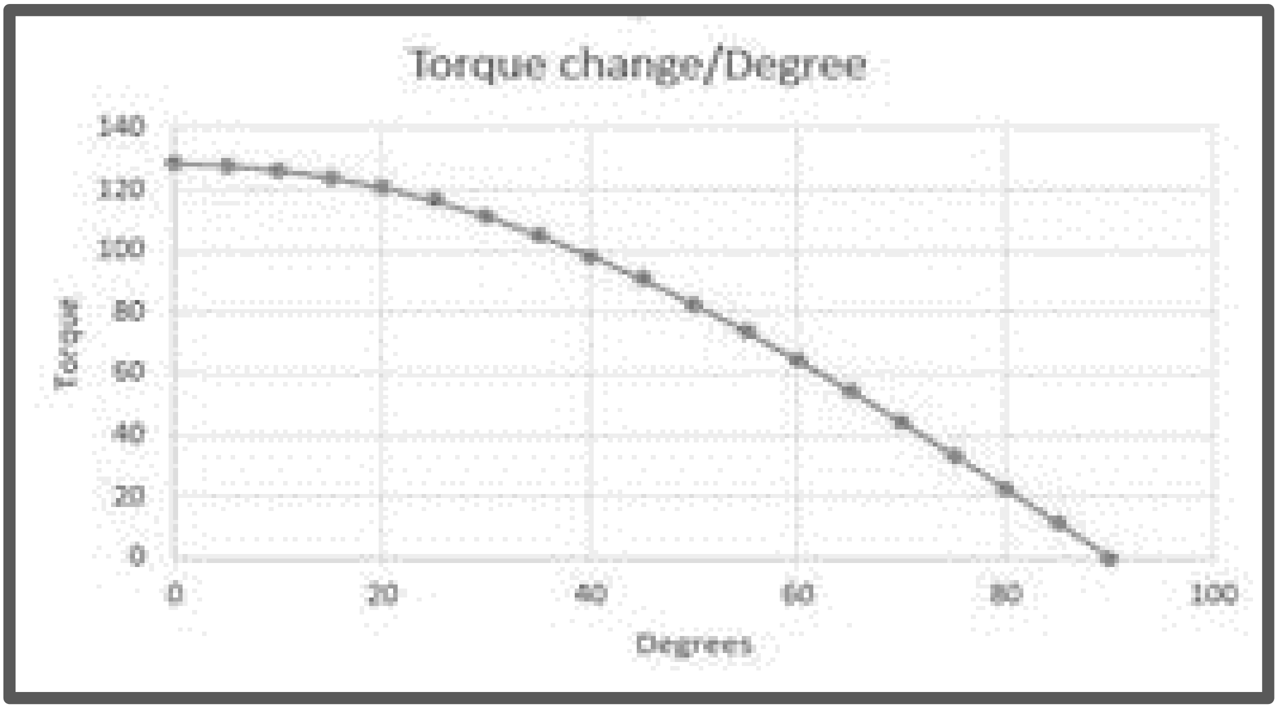


Figure 39 20cm

The above plotted graphs represent the trend of change in angle projection and its effects on the torque calculations for different lengths of bar.

With a change of 5deg eighteen values were taken for each configuration and the plotted in the form of above graphs.

As the angle increases torque requirements decline. But the fact of the matter remains that initial torque requirement is very high in every case and motors that can provide such high torque capabilities are very large in size and require a lot of power for their actuation. Both of which clashed with our design objectives.

This was the reason that conception of a Multi-actuated series of active joints was dropped for the biomechanical system in the initial stages of development and we settled for a passive series of plates and hinge assembly which could be actuated using a mechanism that provides high mechanical advantage.

3.3 Design conclusion

From working on the above design parameters, we concluded the following necessities for the designing of the active part of the device

- There is a need for high torque capabilities
- There is a need for a mechanism that can provide good mechanical advantage
- High load bearing requirements and flexibility

Based on the above design inputs following critical components of the device were selected

- Motors
- Mechanism to be used
- Mechanical supporting structure

3.3.1 Motor selection

To provide actuation for the system a pair of motors were required. Two DC geared motors were selected for this purpose. This selection was based on the following parameters

- Effective operation in the desired application
- Easy to control
- Readily available

These motors had the following specifications

- 300kgcm plus torque providing capacity.
- Power input requirement for the motor were 24volts, 10A
- PWM operated

Find below the pictures of the motors used.

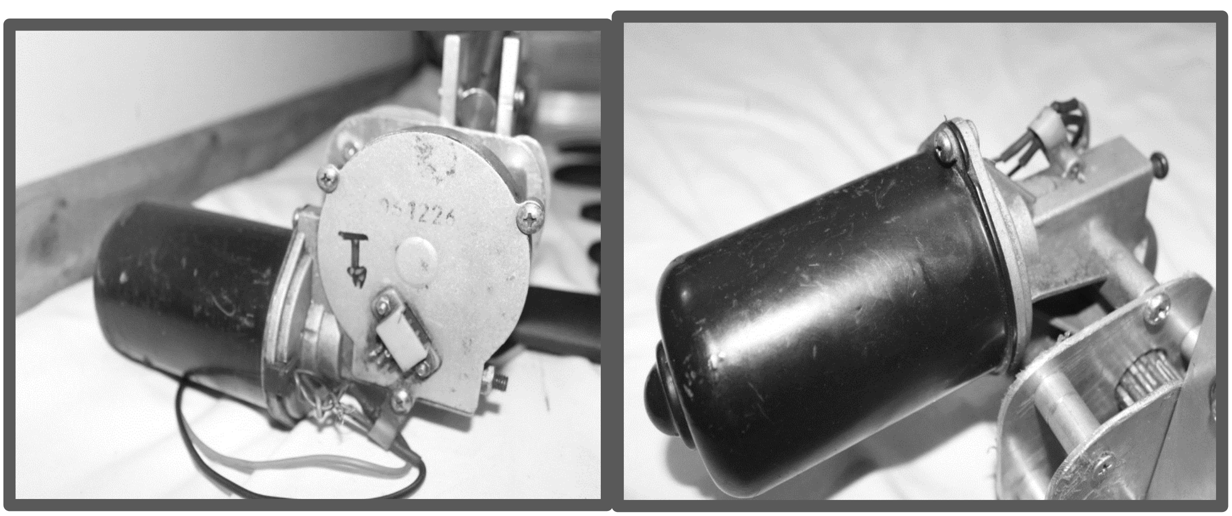


Figure 40 Motors

3.3.2 Selection of mechanism

The basic mechanism used to help with the actuation was “lead screw”. It is an excellent mechanism to convert rotary motion into linear actuation with high mechanical advantage. This mechanism was selected on the following basis

- large load carrying capability
- Compact
- Simple to design
- Easy to manufacture
- **Large mechanical advantage**
- Precise and accurate linear motion
- Smooth, quiet, and low maintenance
- Minimal number of parts
- Self-locking

Moreover, a 4-bar mechanism was used in addition, to provide for the lifting where lead screw worked as a dynamic bar (length changes as actuation occurs).

3.3.3 Mechanical supporting structure

For the mechanical supporting structure, we used a series of metallic plates with passive hinge joints between them. There were three types of plates were used. Each type representing one region of the spine (cervical/thoracic/lumber).

Using this supporting structure provided us with the following desired design requirements

- Load bearing requirements
- Flexibility

This mechanism equips each plate with two D.O.F. Rotation along medio-lateral axis and restricted rotation along antero-posterior axis which allows for the passive assembly to bent around the spine to accommodate for the natural spinal curves. The series of yellow plates and green hinges in the figure below represent the mechanical supporting structure.

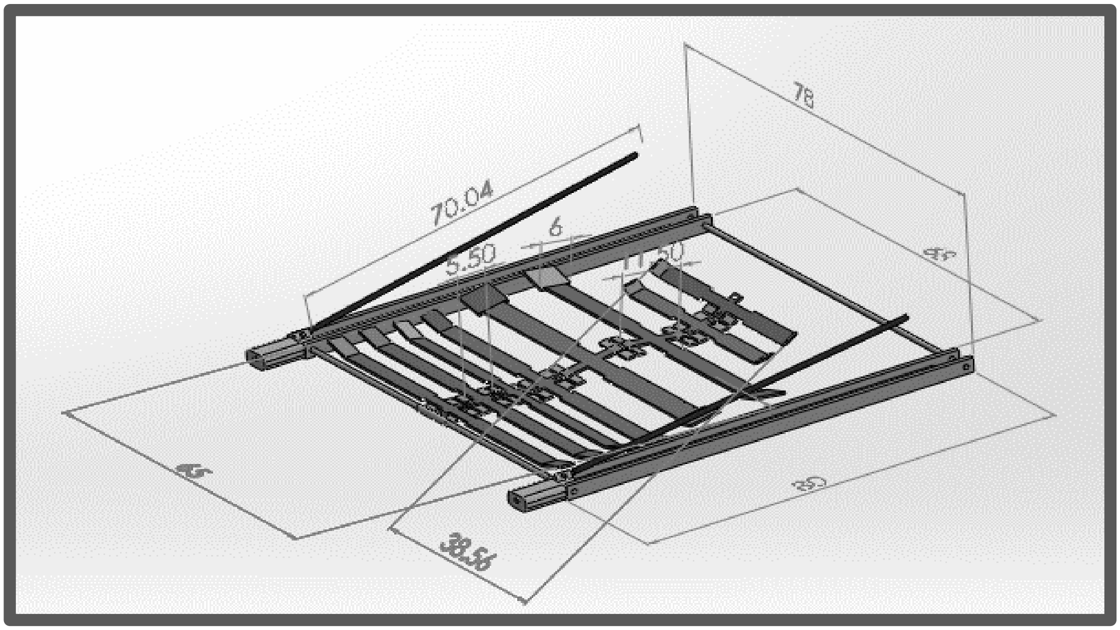


Figure 41 3D Model

3.4 3D modelling and material assignment

In this section we are going to discuss the 3D modeling and material assignment for different components of the mechanism. We are going to start off with showing different views of the complete assembly and then we are going to move forward with individual part dimensioning's and respective material assignment.

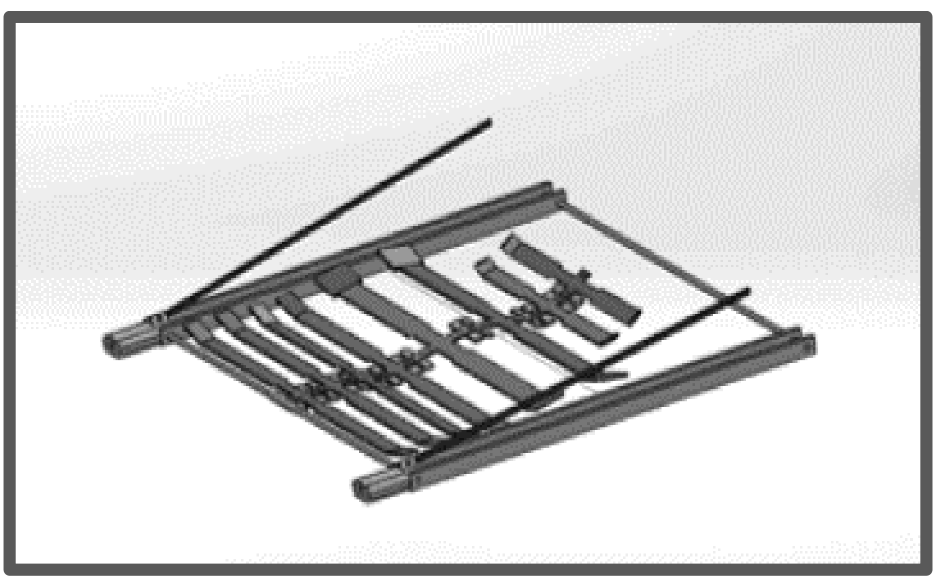


Figure 42 Isometric View

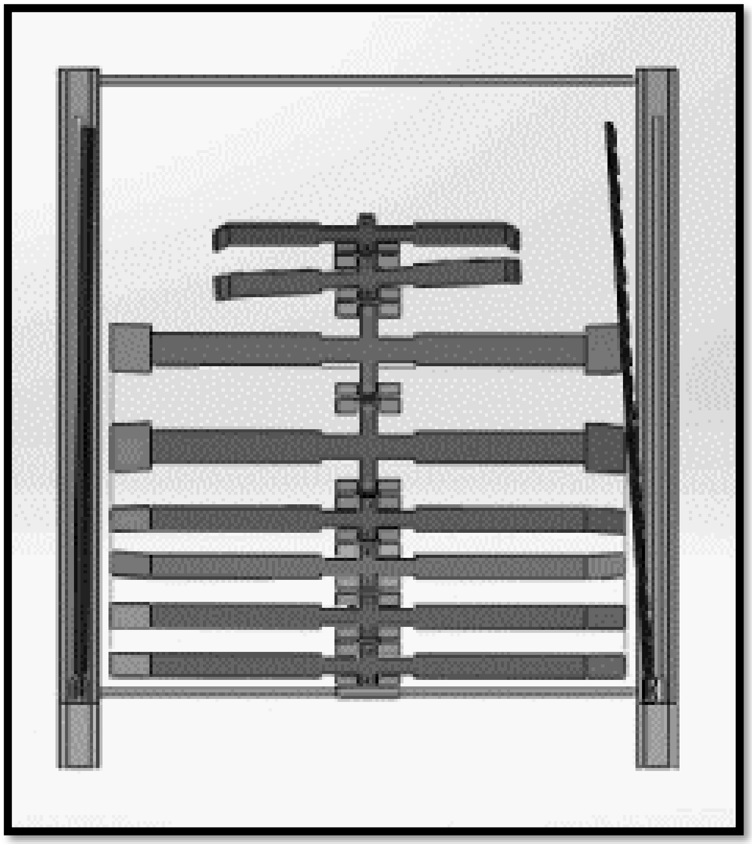


Figure 43 Top View

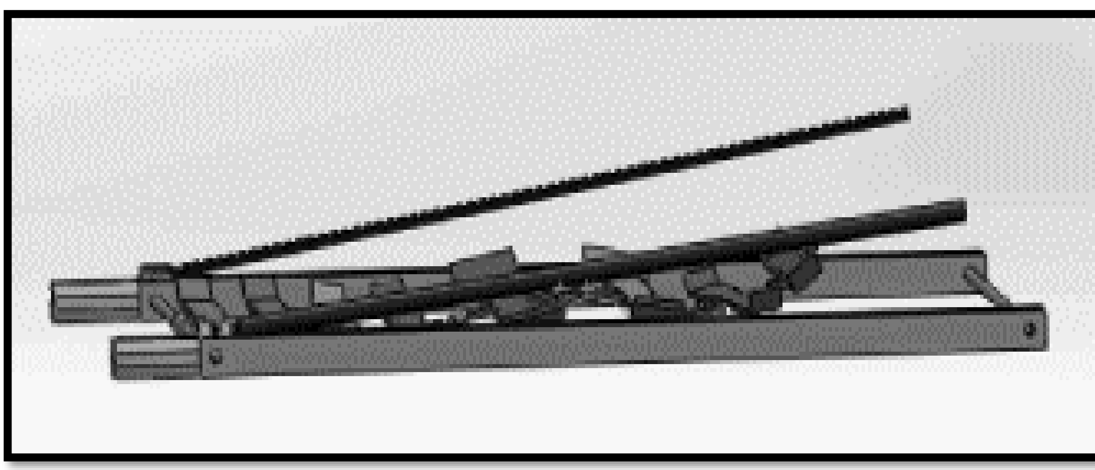


Figure 44 Side View

3.4.1 Cervical plate

- 2 in total
- 25cm/3cm
- 6mm Aluminum

Other dimensions can be seen in the figure below

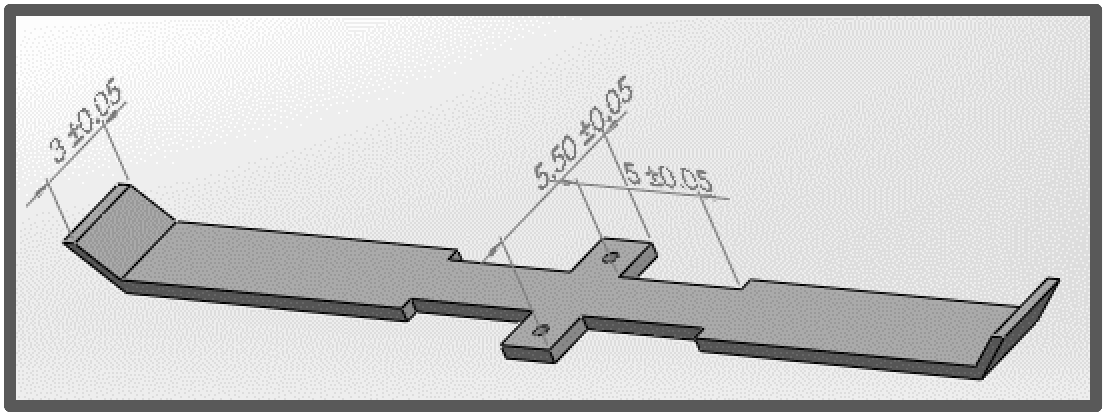


Figure 45 Cervical Plate

3.4.2 Thoracic plate

- 2 in total
- 50cm/4cm
- 6mm Aluminum

Other dimensions can be seen in the figure below

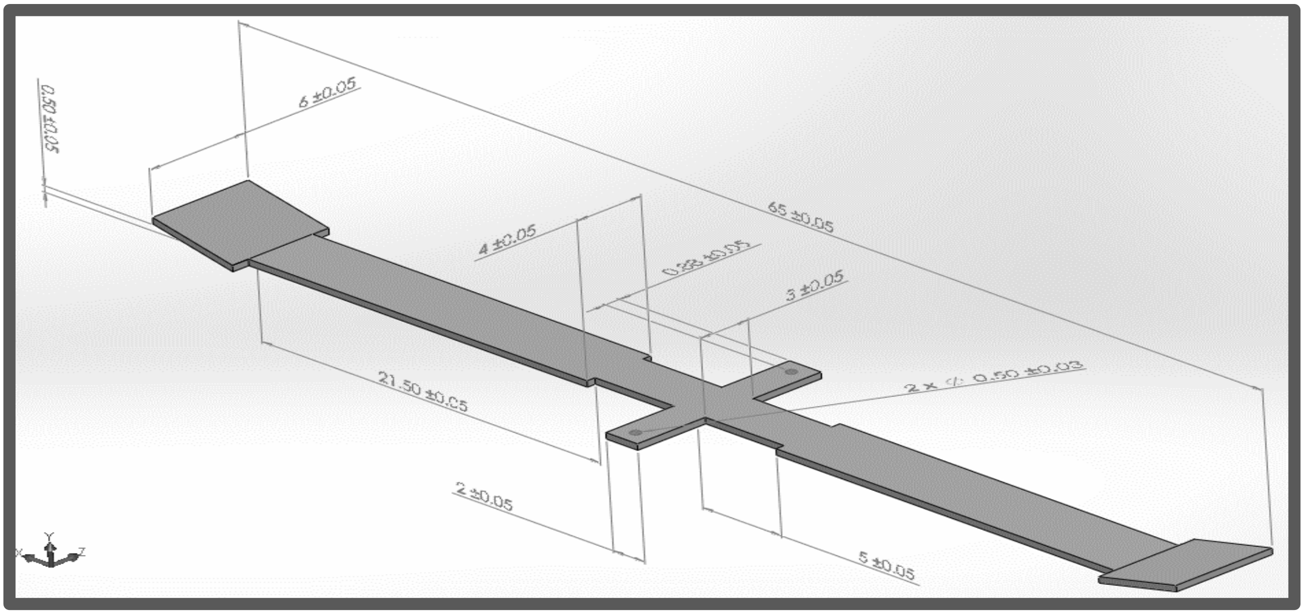


Figure 46 Thoracic Plate

3.4.3 Lumber plate

- 3 in total
- 45cm/3cm
- 6mm Aluminum

Other dimensions can be seen in the figure below

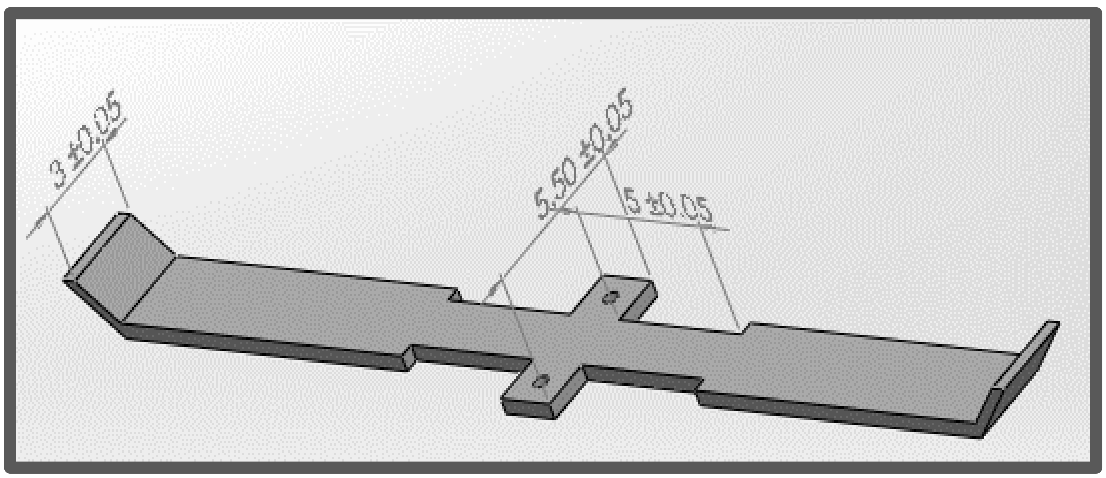


Figure 47 Lumber Plate

3.4.4 Hinge Joint

- Consists of two parts
- 7 pairs in total
- Mild steel

- 4mm Stainless steel wire connects the two parts

Dimensions can be seen from the figures below

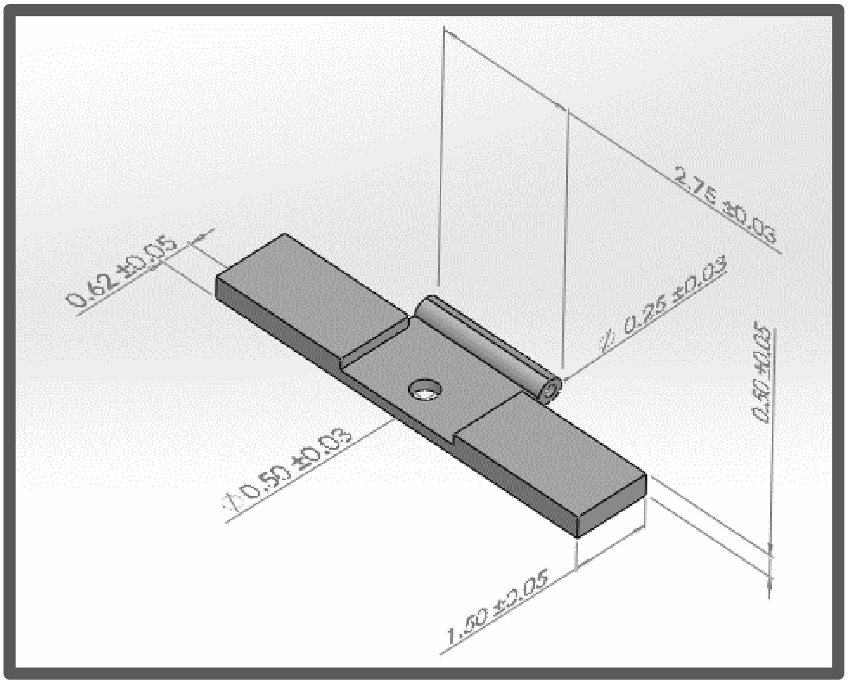


Figure 48 Hinge part 1

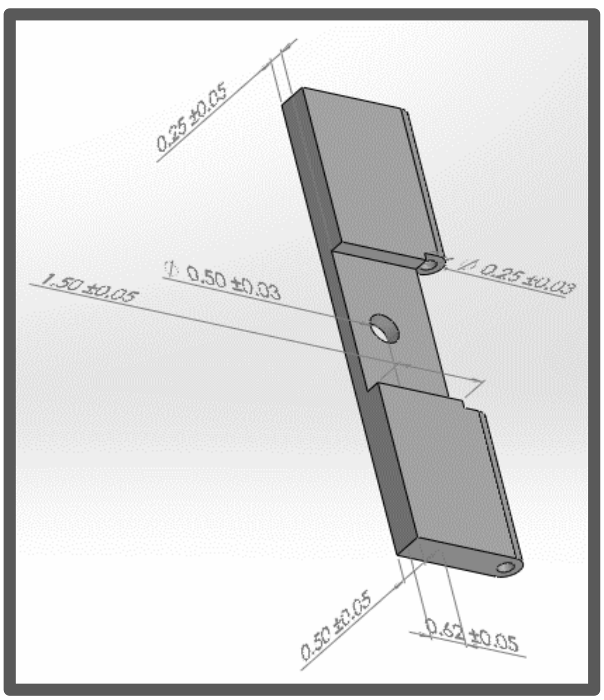


Figure 49 Hinge part 2

3.4.5 Lead screw

- 20mm Diameter
- Pitch 2.5
- 65cm long
- Mild steel

Find below figure showing the modeled lead screw

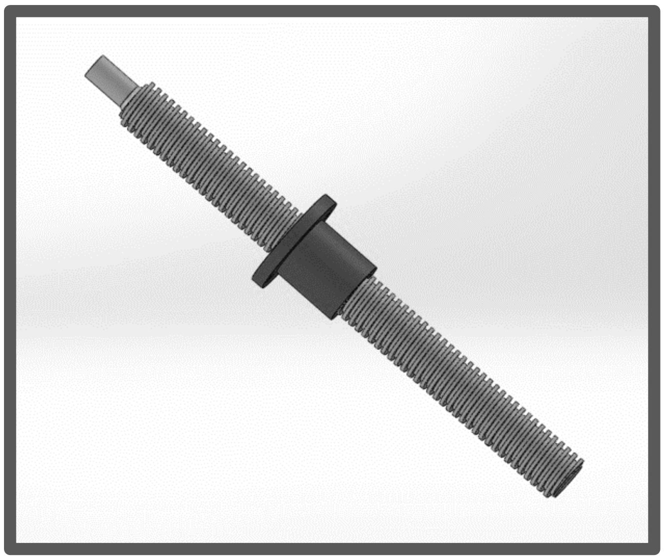


Figure 50 Lead Screw

3.4.6 Pipe and frame

- Pipe
  - Stainless 16 gauge

Find below figure showing the modeled pipe

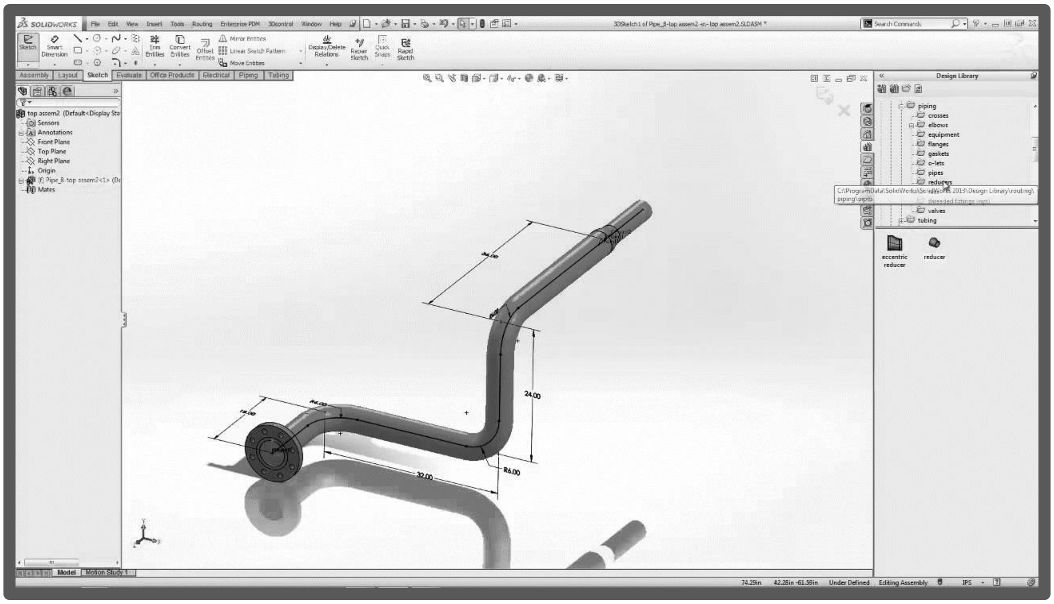


Figure 51 Stainless Steel Pipe

Frame of the device was made from angle iron of aluminum. Specifications of the angle iron were as follow

- 3mm thick
- 1.5" wide

3.4.7 Assembled mechanism

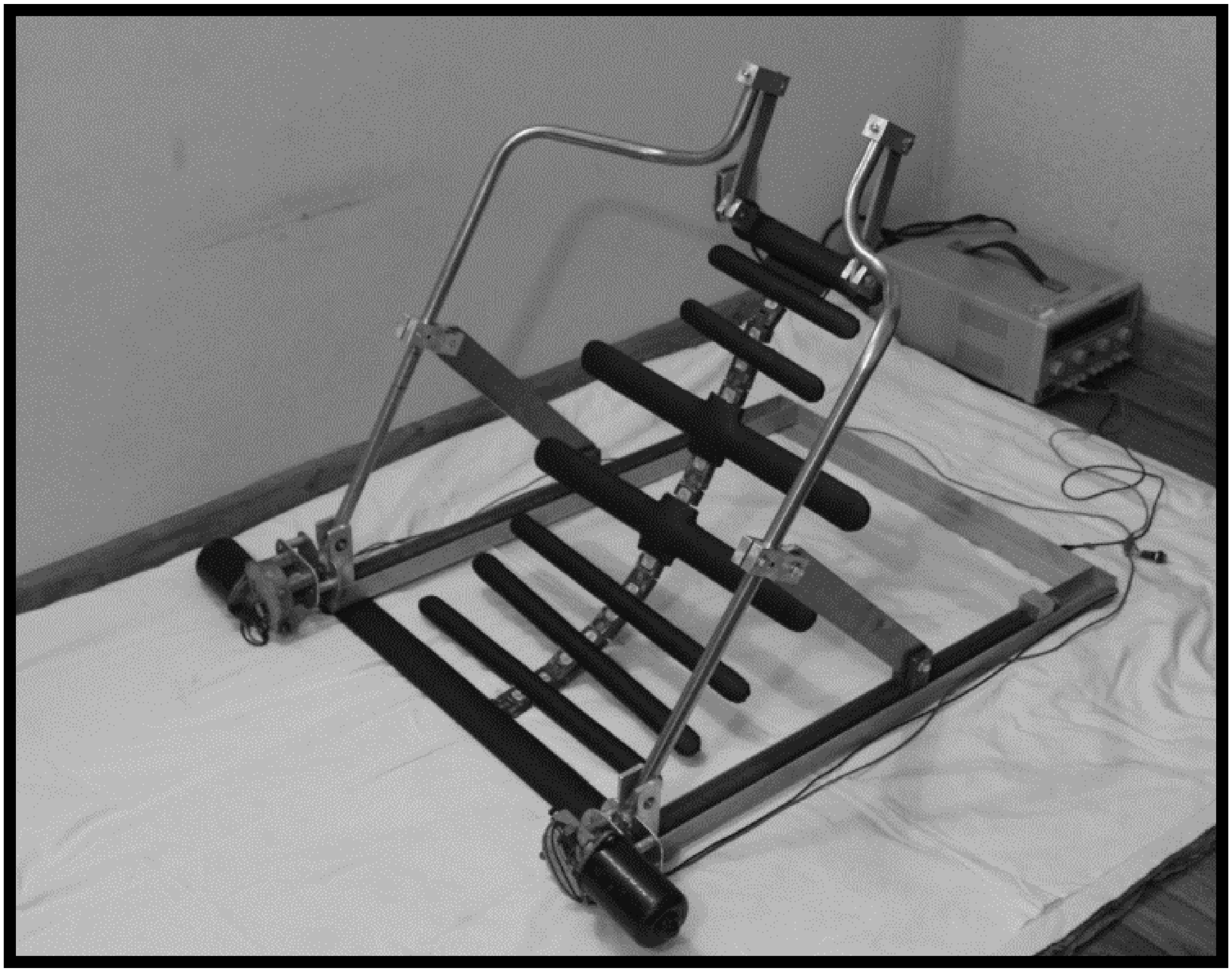


Figure 52 Fabricated Mechanism

The above figure shows the fabricated mechanism and complete assembly of the device. The device requires a power supply which provides the required power for the biomechanical system or orthotic device.

3.4.8 Lifting mechanism

The lifting mechanism composed of a 4-bar mechanism where one of the bars was in the form of lead screw. The working of the mechanism is as follow

- Power supply provides for the required potential for the motors.
- Shaft of each motor is connected to lead screw through two spur gears.
- When the motor is powered on it rotates the lead screw.

- A mild steel block with internal threading linearly translates on each lead screw because of rotation provided by the motor. Converting rotating actuation into linear translation.
- This mild steel block is joined with an aluminum bar which allows for one degree of freedom in the form of rotation.
- The other end of this aluminum link is bolted to another block which can be variably fixed on the stainless steel 16-gauge pipe. This block is made of aluminum.
- The S.S pipe is pivoted at the lower end providing one degree of freedom in the form of rotation in the medio-lateral axis. The other end (top end) is attached to the head rest of the passive assembly through a block and link mechanism.
- Due to pivoting of the pipe and fixation of link (between the pipe and the lead screw), when the mild steel block translates on the lead screw because of rotatory actuation. The pipe is enforced to rotate about the pivot point.
- This allows for the actuation of the passive chain of plates and hinge assembly to impart the desired motion (sitting upright from a lying posture and vice versa).

  Refer to the following figures to grasp the above asserted concept of actuation of the lifting mechanism.

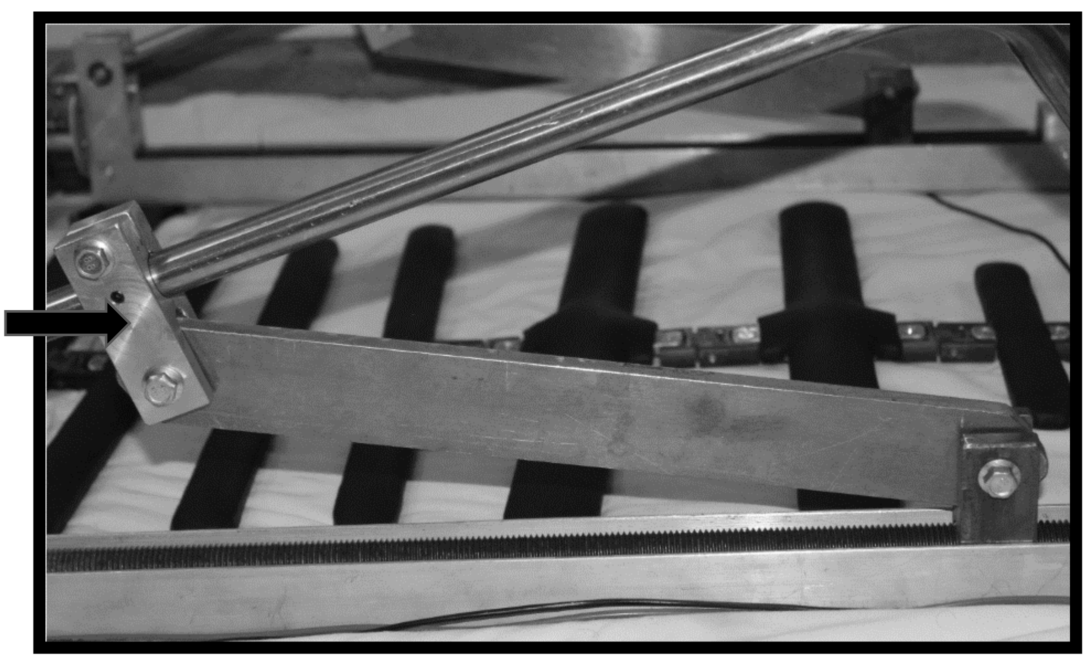


Figure 53 Lateral view of the lifting mechanism

The highlighted aluminum block in the above figure is fixed on the pipe but can be variably adjusted.

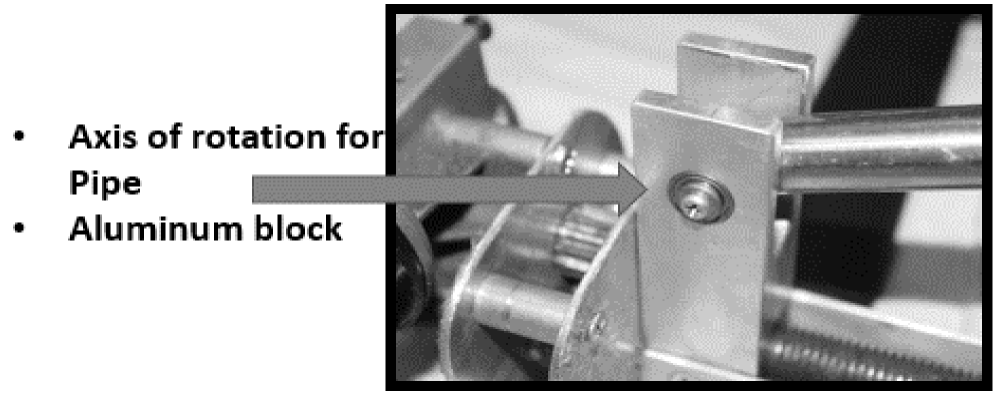


Figure 54 Pivot/fulcrum for the S.S pipe

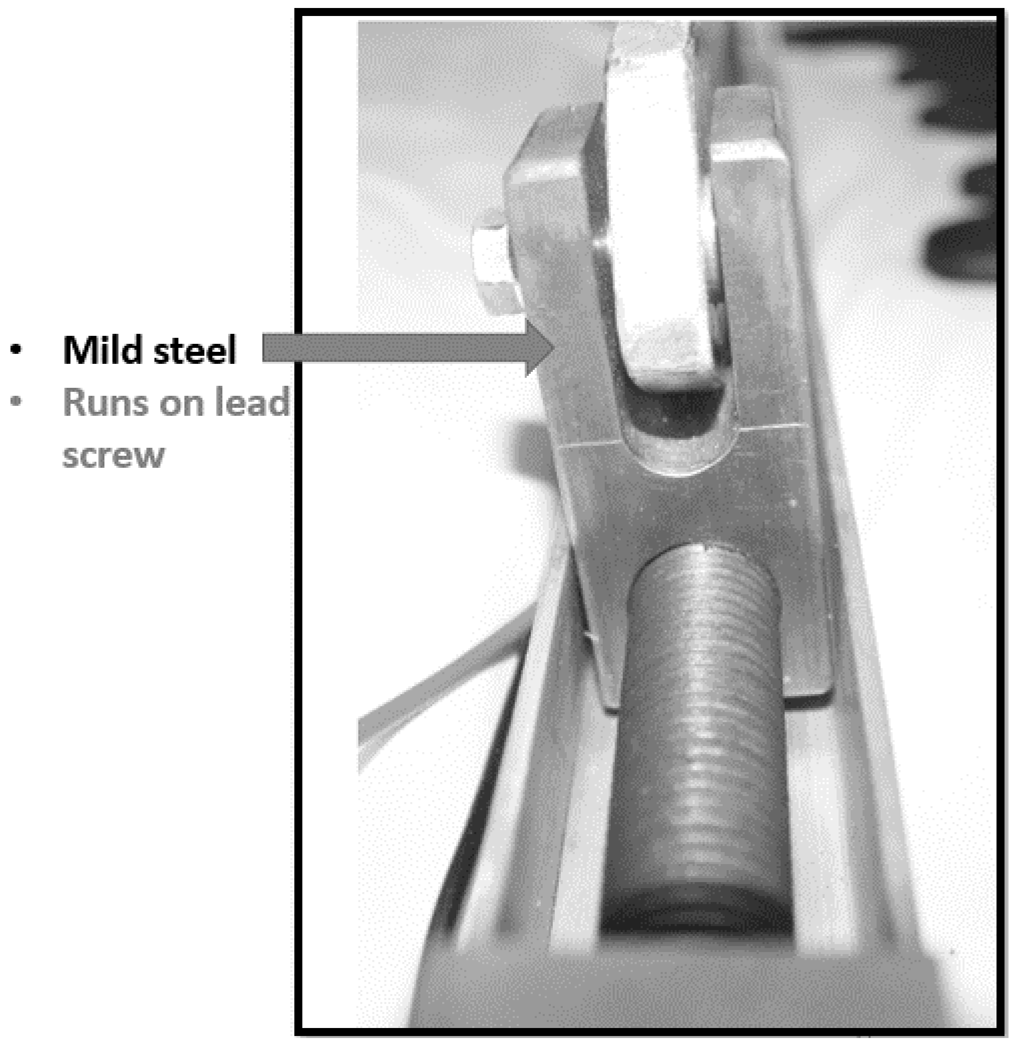


Figure 55 Mild steel block

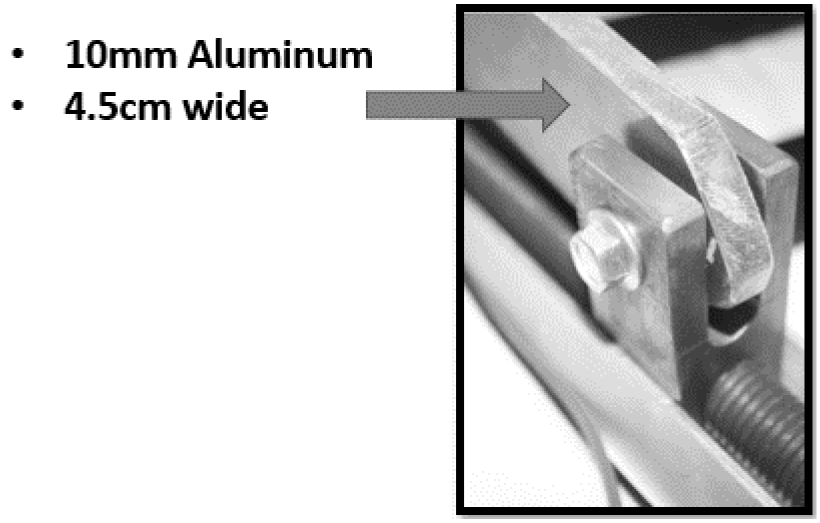


Figure 56 Aluminum Link

3.4.9 Torque capability of Lifting mechanism

Now we are going to see how much torque can be provided by the lifting mechanism. This is an essential step. For we need to make sure that the torque capabilities are more than the torque requirements for the biomechanical system or the orthotic device [30].

We know each motor provides more than 300kgcm torque. As two motors re being used, we will get 600kgcm plus torque from only the two motors.

We also are aware of great mechanical advantage provided by the lead screw mechanism. Even though such mechanism provides a very high mechanical advantage in theory. A significant part of this advantage is lost in overcoming the friction in real life applications. So, for this reason we only considered M.A provided by lead screw to be 3.

Based on the above assumption following calculations were made

Torque of motor= 300kgs

Torque after Lead screw= 3*300=900kgs

2 motors= 2*900=1800kgs

Required torque=1545kgs

Torque provided by the lifting mechanism is more than the required torque by 255kgs. Even though we achieved torque more than the requirements. We still wanted to better the torque capabilities of the biomechanical system. So, to increase this difference (between the provided torque and required torque) a pair of spur gears were used between the motor shaft and lead screw [30].

The pair of gears was used following the torque enhancer configuration. Which means one of the gears was smaller than the second one. The smaller gear was set to be the driver and the bigger one as the follower. This increased the torque even more. Refer to the following figure to grasp the above asserted concept.

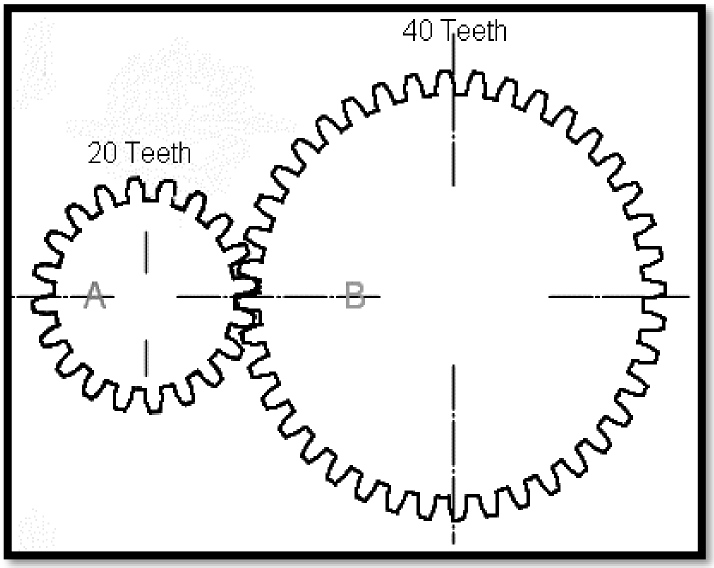


Figure 57 Torque enhancer configuration

Gear A is the driver and Gear B is the follower. This configuration increases the torque but decreases the speed. See the following figure pertaining to incorporation of the torque enhancer configuration (of the gears) in the biomechanical system.

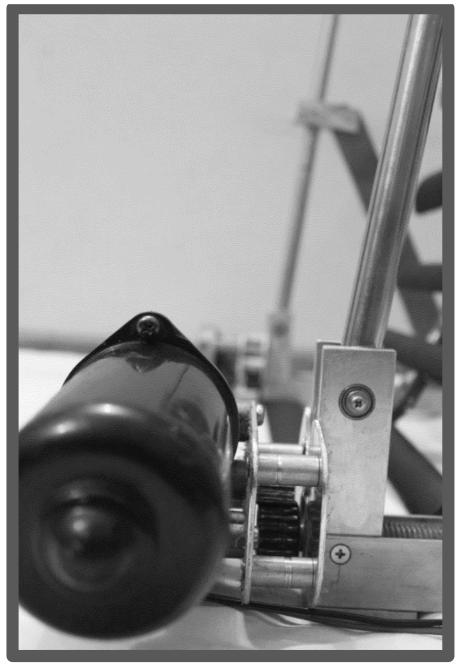


Figure 58 Torque enhancer incorporated

CHAPTER 4 Results and Discussion

4.1 Initial testing

Once the mechanism was assembled, it was time to test the mechanism of its working. Initially, we had to test the mechanism for a complete posture change without any load on it. This was carried out to ensure that the mechanism didn't jam at any point of the motion and to pick any issues pertaining to the achievement of desired movement.

Refer to the following collage below to see initial testing of the device

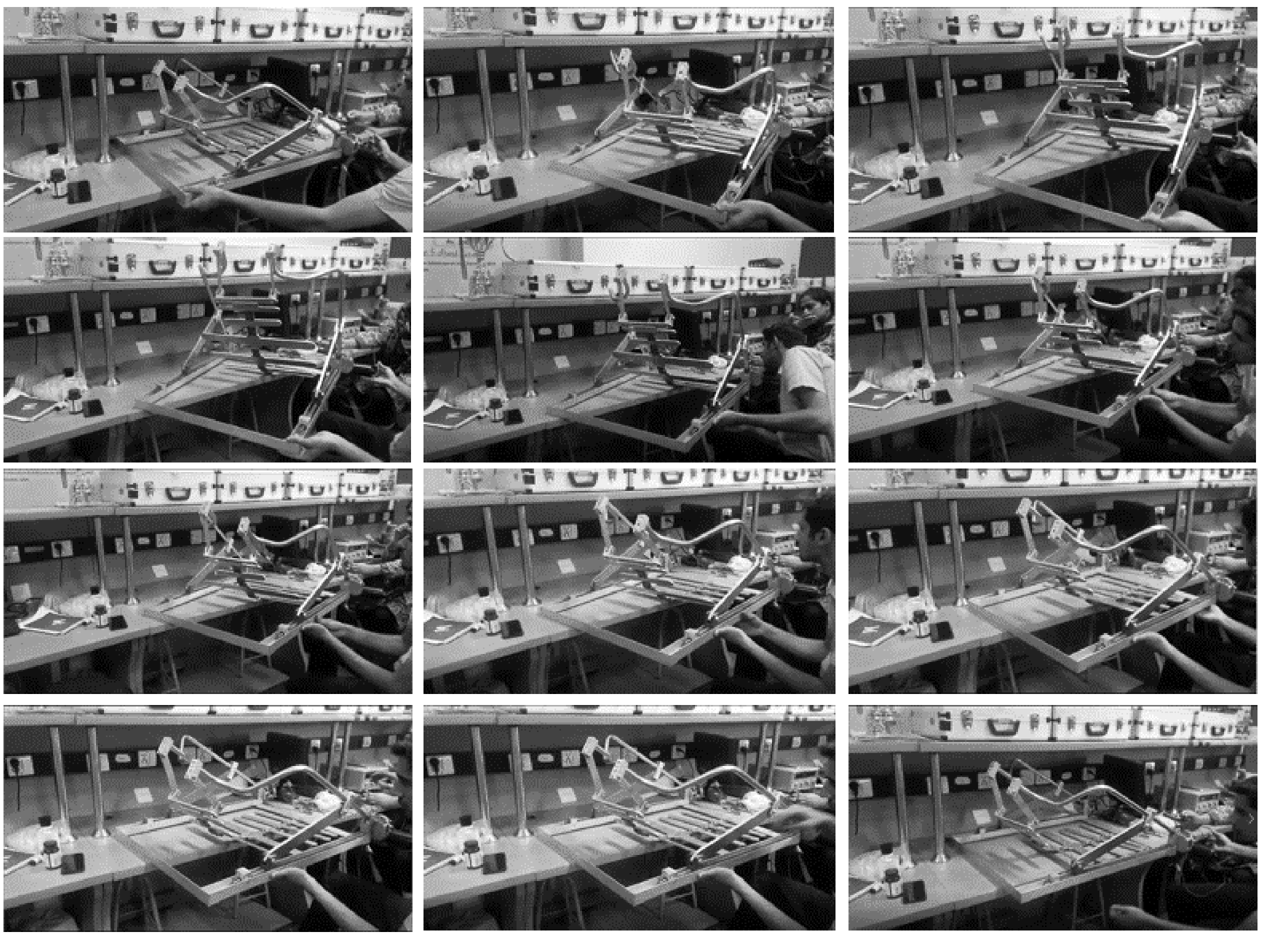


Figure 59 Initial testing of the device with no load

4.2 Issues faced in Initial testing

4.2.1 Motor mismatch

As we know the system comprised of two motors, each actuating a lead screw. We were bound to face the issue of motor mismatch. Motor mismatch basically means that one motor works/rotates sluggishly compared to the other and that makes it difficult for uniform actuation of both arms of the system. This issue is usually rectified digitally using a microcontroller which reads the rotation of both motors and makes them rotate at the same pace solving the issue of mismatch. But to do this we need to have some sort of an equipment to read the rotation of motors.

Here we introduce the concept of motor encoders. Motor encoders are used to read the rotation of the motors. There are two main types of encoders, namely

- Optical encoders
- Rotary encoders

For the device we used optical type. This type comprises of two parts.

- Encoder disk
- Encoder module

Encoder disk is just a disk with uniformly distanced slits on it which is fixated on the motor shaft and rotates to the same degree as the motor shaft. Whereas the encoder module reads these slits through an assembly of IR transmitter and receiver and gives the respective response in the form of TICKs to the microcontroller which corresponds to them accordingly.

For the device we did not place the encoder disks on motor shafts. Rather we fixated the disks at the other end of the lead screws and encoder modules were placed to be able to sandwich these disks, allowing us to read the TICKs. This does not affect the proceedings of correcting the mis match as it is done on both sides of the assembly and since shafts of both motors are connected directly to the lead screw through a spur gear assembly with same specifications.

Refer to the following figure pertaining to motor encoders

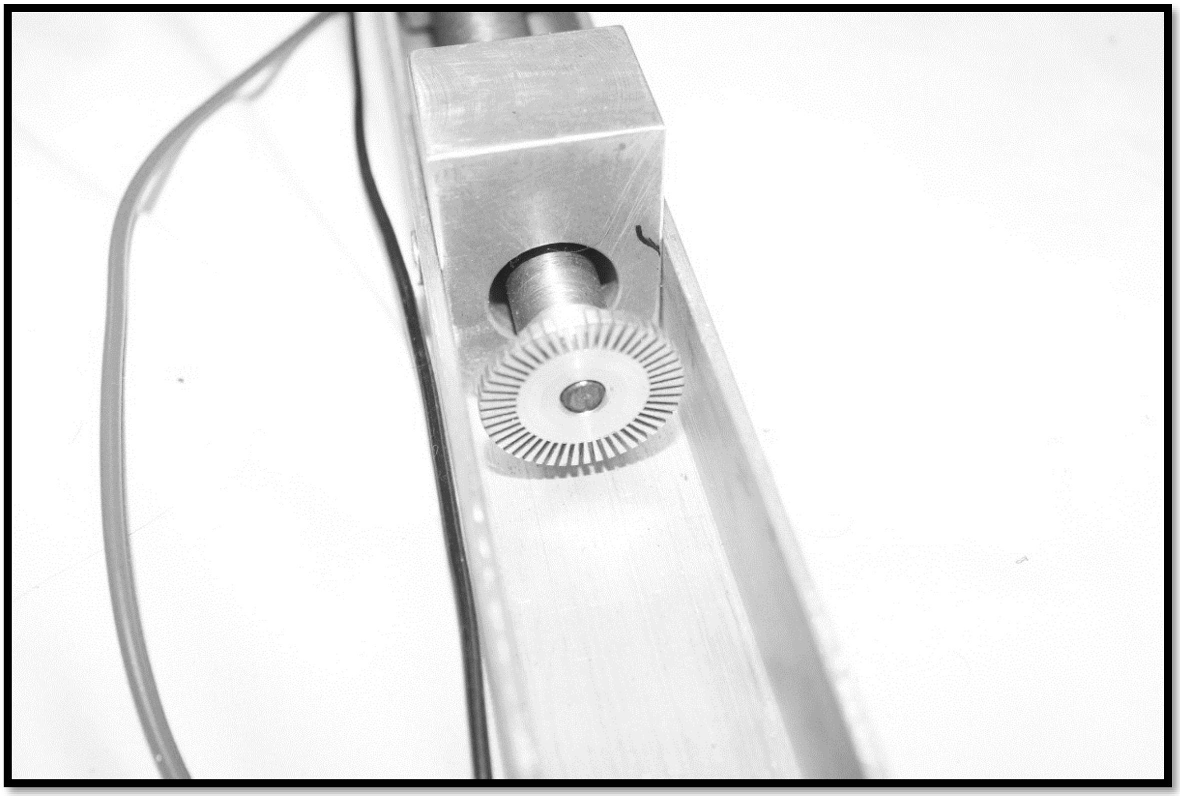


Figure 60 Encoder disk with 48 slits giving us a total of 96 TICKs per revolution

4.2.2 Padding issue

Since we used metallic plates and hinges to cover the human back in ours design. Initially, when we tested the machine with patients. They complained about the plates to be too rigid. Highlighting the need of padding. So, all the plates were padded using a foaming pad of 1.5-inch depth and then covered with latherite material to hold the padding and to better the esthetics of the device.

4.3 Testing with Patients

Find below the collage of figures which comment on different stages of the desired postural changes.

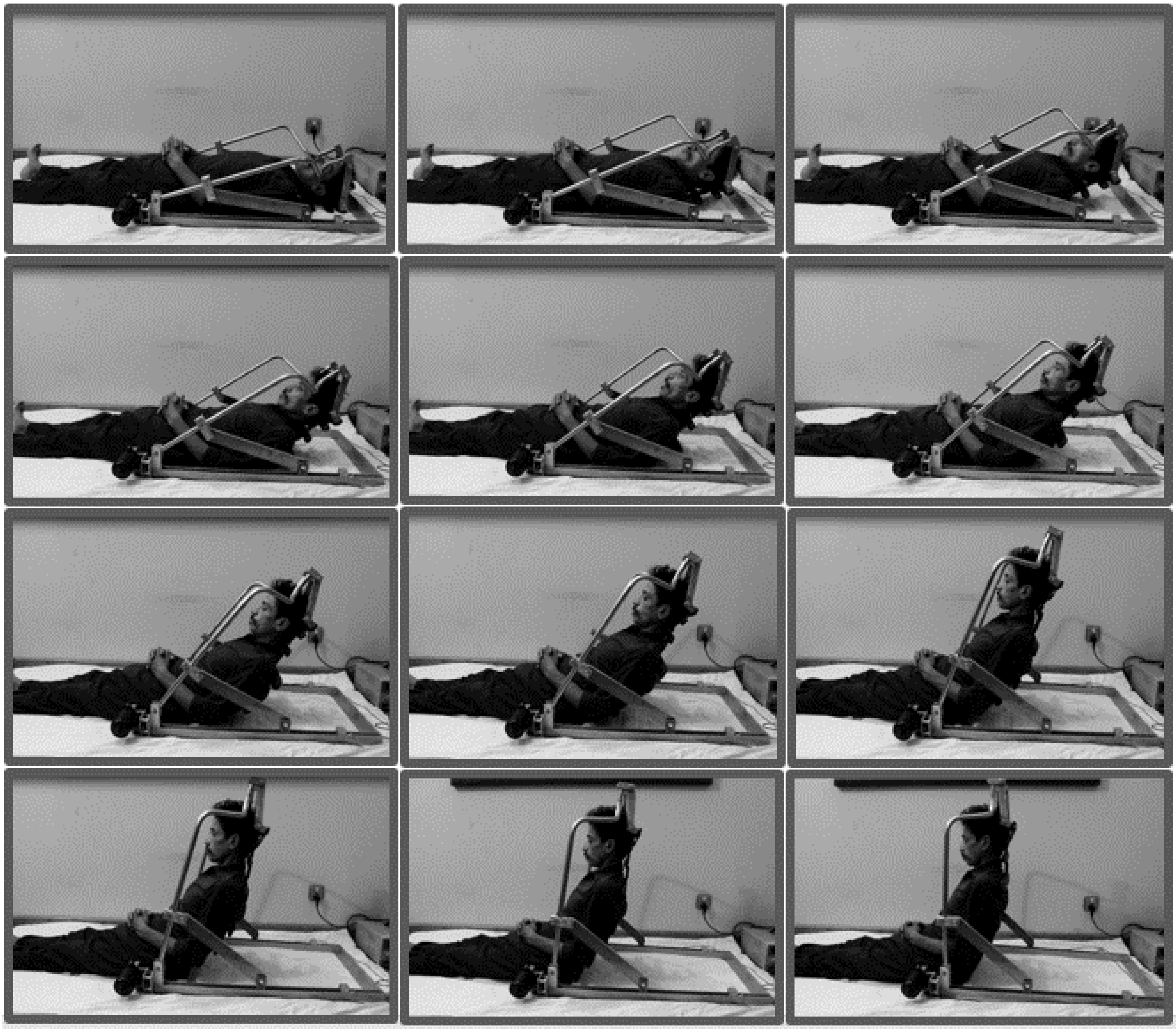


Figure 61 Different stages of the desired postural changes

It can be clearly seen from the above series of figures that the made device not only achieves both extreme desired positions (sitting up and lying down) but also follows biomechanics of spine in doing so.

When we say the device is capable of following biomechanics of spine. It is based on the following observations.

- When a human body sits up from a laying posture. In the initial stages of the posture change the head lifts and the cervical region provides for such lift by rotating about the upper thoracic region.
- After the initial stages the thoracic region of the spinal column lifts itself making a curl with the lumber region which is the lower back and is still parallel to the horizontal surface it is placed upon. These are the Mid stages of the postural change.
- In the final stages the lumber region lifts and pivots at the pelvic girdle to achieve an upright posture. During the final stages, lumber region and cervical regions realign to minimize the curl created while the body was lifting.

From the above collage of figures, it can be clearly seen that the device allows for the desired postural change while respecting the stages of such change of posture, namely

- Initial stages
- Mid stages
- Final stages

Hence, we can say that the device has been able to provide for the required actuation and respecting the inherent biomechanics of spine in doing so.

4.3 Results and conclusion

It is evident from the above section that we have been able to achieve all the research objectives that we set for ourselves.

We have been able to come up with a biomechanical palliating system that allows for bed ridden patients to sit from a laying posture and vice versa while respecting the biomechanics involved. Also, we were able to design the device in a way that it allows for comfortable postural changes.

We have made the device user friendly. It is safe to operate. It is easily equipped and is very easy to operate.

But most importantly we were able to come up with a solution that allowed for such postural changes and was at the same time portable. The solution or device can be placed on any bed and give that bed the capability of allowing for such postural changes. This is one of the greatest novelties of this device along with the passive assembly made to mimic the biomechanics of spine.

4.4 Comparison with adjustable bed

How is it a better solution?

PARAMETERS	ADJUSTABLE BED	OUR PROTOTYPE
Cost	1200-2200	Low cost: 570$
Size	Large	Smaller
Weight	Heavy	Light
Biomechanics	Does not mimic	mimics
Portability	Low: needs more than one person to move	High: Can be moved by one person

Novelty: it can be placed on any bed to turn it into a better adjustable bed.

4.5 Advantages and Other uses

Apart from the set research objectives the device it can be used for the following advantages and uses

- It can be used to cater for Bed Sores:
  - Weight shifted once a day can stop bed/pressure sores
- It can be used for Physiotherapy:
  - Muscle atrophy treatment
  - Muscle strengthening
- It can be used as a rehabilitative (ROF) and Assistive device

4.6 Future works

- Controls to be applied
- Device can be integrated with EMG
  - Will better the interaction of patient with the environment (giving control to the patient)
- Can be used for postural studies using pressure sensors.

CHAPTER 5

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