

HUMAN SPINE KINEMATICS FOR APPLICATIONS IN
MULTIPLE-JOINT SPINAL ORTHOSIS

WAN ALIFF BIN ABDUL SAAD

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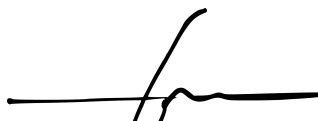
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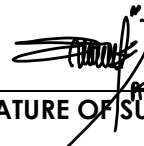
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PKM 163009

MATRIC NUMBER



SIGNATURE OF SUPERVISOR

DR MOHD AZUWAN
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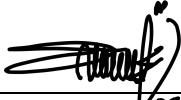
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Signature : 
Name of Supervisor I : MOHD AZUWAN MAT DZAHIR
Date : 6 OCTOBER 2021

Signature : 
Name of Supervisor II : MOHAMED HUSSEIN
Date : 6 OCTOBER 2021

Signature : 
Name of Supervisor III : MAZIÁH MOHAMAD
Date : 6 OCTOBER 2021

Signature : 
Name of Supervisor III : SHAHARIL MAD SAAD
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HUMAN SPINE KINEMATICS FOR APPLICATIONS IN
MULTIPLE-JOINT SPINAL ORTHOSIS

WAN ALIFF BIN ABDUL SAAD

A thesis submitted in fulfilment of the
requirements for the award of the degree of
Doctor of Philosophy

School of Mechanical Engineering
Faculty of Engineering
Universiti Teknologi Malaysia

OCTOBER 2021

DECLARATION

I declare that this thesis entitled “*Human spine kinematics for the application of multiple-joint spinal orthosis*” is the result of my own research except as cited in the references. The thesis has not been accepted for any degree and is not concurrently submitted in candidature of any other degree.

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DEDICATION

This thesis is dedicated to my mother, who taught me that the best kind of knowledge to have is that which is learned for its own sake. It is also dedicated to my wife, who taught me that even the largest task can be accomplished if it is done one step at a time. Also, to both of my princesses that have taught me to be a better person as each beautiful day passes.

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ABSTRACT

Low back pain (LBP) treatments necessitate a full understanding of spinal kinematics since the spine is made up of many segments. However, there is still a limited study that focuses on the active and passive spinal segments during critical daily activities. Even data profile on the spinal multi-segment contribution to spinal kinematics has not yet been established. Currently, there is no intelligent multiple-joint spinal orthosis that emphasizes segmental biomechanics of, and provides assistive support to the spine. Most of the literature studies only focused on up to three joints or segments mechanism. The main objective of this research was to analyze the postural movement of spinal motion in critical daily activities. This study also focused on the active-passive spinal segments of the thoracic and lumbar, as well as their contributions to spinal motions. This study was divided into three phases, with the first focusing on a fundamental study of critical daily activity, the second on product design development and control system of the proposed multi-joint spinal orthosis, and the last on the integration of the control system, prototype, and data profile. The interclass correlation coefficient (ICC) analysis showed that all tasks investigated had excellent reliability, with values ranging from 0.94 to 0.99. In both the Begin and Return cycles, two-way analysis of variance (ANOVA) demonstrated significant differences in findings for interaction between spinal segments and different approaches within thoracic segments ($P\text{-value} < 0.05$). The interaction within the lumbar region was found for Weightlift 5kg load during the Begin cycle and in both cycles during the Pick Object task. For both weights lifted, it was discovered that the Stoop approach produced a peak of angular displacement that was about 4 times greater than the Squat. A motion capture device was used to obtain the data profile for the active-passive spinal segment. The active-passive range was chosen based on electromyography activation data from similar studies in the literature. A prototype of the multi-joint spinal orthosis was developed together with the initial integrated mechatronic system. This study has successfully contributed to a data profile of active-passive spinal segments, which is critical in comprehending complex spinal kinematics.

ABSTRAK

Rawatan bagi sakit belakang (LBP) memerlukan pemahaman penuh mengenai kinematik tulang belakang yang terdiri daripada beberapa ruas. Namun, sangat sedikit kajian yang menumpukan pada ruas tulang belakang aktif-pasif semasa melakukan aktiviti harian. Malah, susuk data berkaitan sumbangan berbilang ruas kepada kinematik tulang belakang belum dapat dipastikan. Pada masa ini, tidak ada ortosis tulang belakang berbilang ruas yang memiliki kecerdikan yang menekankan biomekanik tulang belakang yang beruas dan memberikan sokongan bertujuan membantu tulang belakang. Sebilangan besar kajian literatur hanya menekankan pada mekanisme tiga sendi atau ruas. Objektif utama penyelidikan ini adalah untuk menganalisis pergerakan postur bagi tulang belakang dalam kegiatan harian yang kritikal. Kajian ini juga memfokuskan pada ruas tulang belakang aktif-pasif bagi toraks dan lumbar, serta sumbangannya terhadap pergerakan tulang belakang. Kajian ini dibahagikan kepada tiga fasa, dengan fasa pertama memfokuskan pada kajian asasi bagi kegiatan harian yang kritikal, fasa kedua mengenai pembangunan rekabentuk produk dan sistem kawalan bagi ortosis tulang belakang berbilang ruas yang dicadangkan, dan fasa terakhir mengenai integrasi sistem kawalan, prototaip, dan susuk data. Pekali sekaitan antara kelas (ICC) menunjukkan semua tugas mempunyai keboleharapan yang tinggi, dengan nilai antara 0.94 hingga 0.99. Dalam kitar Mula dan Kembali, analisis varian dua-hala menunjukkan perbezaan yang signifikan bagi interaksi antara ruas tulang belakang dan pendekatan yang digunakan dalam segmen toraks (nilai $P < 0.05$). Interaksi dalam lumbar bagi Angkat Berat 5kg diperolehi semasa kitar Mula dan dalam kedua-dua kitar semasa tugas Ambil Objek. Untuk kedua-dua berat yang diangkat, didapati bahawa Membongkok menghasilkan puncak sudut sesaran kira-kira 4 kali lebih besar daripada Mencangkung. Peranti merakam pergerakan digunakan untuk mendapatkan susuk data untuk ruas tulang belakang aktif-pasif. Julat aktif-pasif dipilih berdasarkan data pengaktifan elektromiografi dari kajian serupa dalam literatur. Prototaip ortosis tulang belakang berbilang ruas telah dibangunkan bersama sistem mekatronik bersepadu pada peringkat permulaan. Kajian ini berjaya menyumbang kepada susuk data bagi ruas tulang belakang aktif-pasif, yang mana sangat penting dalam memahami kinematik tulang belakang yang kompleks.

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LIST OF ABBREVIATIONS

BMD	-	Bone Mineral Density
BMI	-	Body Mass Index
CG	-	Center of Gravity
CMS	-	Coordinate Measuring System,
CNS	-	Central Nervous System
CPG	-	Clinical Practice Guideline
D	-	Dynamic/flexible orthosis
DOF	-	Degree of freedom
FEA	-	Finite Element Analysis
HH	-	Hanging Hand
IC	-	Iliac Crests
kg	-	kilogram
kN	-	Kilo Newton
LBP	-	Low Back Pain
LD	-	Left Direction
LLA	-	Lower Lumbar Angle
LTA	-	Lower Thoracic Curve
MP	-	Motion Palpation
MRI	-	Magnetic Radio Imaging
MTA	-	Middle Thoracic Angle
N/A	-	Not Available
O	-	Old Subjects
PDS	-	Product Design Specifications
PSIS	-	Posterior Superior Iliac Spines
R	-	Rigid
RD	-	Right Direction
ROMs	-	Range of Motions
RSS	-	Root of the Spine of Scapula
SB	-	Straight Back
SH	-	Straight Hand

SR	-	Semi-rigid
ULA	-	Upper Lumbar Angle
UTA	-	Upper Thoracic Curve
W	-	With
W/O	-	Without
W/s	-	Weightage Score
WHO	-	World Health Organization
Y	-	Young Subjects
yo	-	Years Old

LIST OF SYMBOLS

%	Percent
°	Degree
>	More than
<	Less than
±	Plus-minus
Δ	Difference

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CHAPTER 1

INTRODUCTION

1.1 Problem Background

Bones are architectural masterpieces. They consist of living tissue, and during a human life span, bones experience the balanced process of breakdown and formation which is known as “remodeling”. The spine, also known as vertebrae, is one of the most important bones in the human body. The vertebrae's primary roles are to support the upper body's weight and to house the spinal cord, which is a nerve system structure that transmits signals that control body movement and convey the feeling. Lower back pain is a term used to describe problems with the spine (LBP). The majority of LBP cases are due to non-specific causes, and most resolve spontaneously [1]. On the other hand, it could be a sign of major problems with the lower spine which are also referred to as red flags. Most notably, red flags are an indicator of critical causes such as cancer, infection, fracture, severe spinal nerve compression, and aortic aneurysm [2]. The main factors leading to LBP are (1) bad posture implemented in daily living, (2) excessive and frequent weight lifted at home or workplace, and (3) high or excessive spinal loads, especially for the after-spine surgery patients. It may lead to implant subsidence, pedicle screw loosening, or even implant failure, and may also be a reason for LBP [3]. Literature also concludes that the instability or loss in the normal pattern of spinal motion is another leading cause of LBP [4]. White and Panjabi defined clinical instability of the spine as the loss of the spine's ability to maintain its patterns of displacement under physiologic loads [5]. A study by Kwok, Leung [6] found that the pervasiveness of LBP that lead to vertebral fracture was due to low hip bone mineral density (BMD), previous long-term occupation, or job, and poor physical condition which is common for the elderly.

LBP is a painful condition that is seen in both general and specialist practice. Many studies have been done to find solutions to this problem, which has impacted everyday human activities such as working environments. It has been estimated that over 20% of all private sector injuries involve the back and in 2009, sprains and strains accounted for 40% of injuries and illnesses resulting in days away from work and most often involved the back. Works of literature found that LBP is a major problem throughout the world, and is common for females and people aged between 40 to 80 years old. In Malaysia, the percentage of LBP rate differs based on geographical population and type of occupation. There are many problems faced by those who suffer from LBP. For example, LBP problems will limit the movement of a person and reduce motion efficiency. LBP can be more than just physical as it causes physiological stress to the patients. In Malaysia, it affects peoples' workdays and an organization's productivity, leading to financial loss.

The "spinal orthosis" is a device that aims to provide solutions for back pain patients when applied to the body to limit the motion, correct deformity, reduce axial loading, or improve the function of a particular spinal segment. Existing passive or unpowered spinal orthosis products can be divided into three different classes, which are rigid orthosis, semi-rigid, and flexible orthosis. Flexible spinal orthotics are constructed out of strong fabrics or elastic materials with a variety of stiffer supports. Rigid spinal orthotics are used when greater control of motion or posture is required. This type of orthosis is fabricated using high-temperature thermoplastics or lightweight metals and has many varieties with a broad selection of pads and coverings. Research has been carried out on active and powered spinal orthosis. However, most of these orthoses were focusing on the single joint, rigid type of orthosis that provides external force or torque on the specified region of the spine. In addition, the control mechanism for the spinal orthosis had not yet been completely developed with most of them still in the research stage. To the best of the author's knowledge, a multiple-joint spinal orthosis that emphasizes the design of the human spine biomechanics is yet to be extensively investigated, then made commercially available. Patients that suffer a back injury (especially related to back pain) usually having difficulty in doing movements involving spinal motion such as forward flexion,

extension, and lateral bending on both sides. Therefore, a spinal orthotic device that can give support to the spinal motion is critically required.

1.2 Problem Statement

The LBP issue is a major cause of morbidity around the world. The solution strategies require depth understanding of the spinal kinematics as the spine is made up of many segments. However, there is still limited study that highlights the active and passive spinal segments during critical daily activities. Thus, a fundamental study related to posture detection on active-passive spinal segments during critical daily activities should be explored. A review of the articles published in the last 10 years shows that around 14 prototypes of powered and active spinal orthoses had been developed. Most of the spinal orthosis only implemented up to two joints/segments mechanism to support patients with back pain and others were implementing different mechanisms. These orthoses had implemented different systems, actuators, sensors, and control systems to produce the best spinal orthosis that gives support for those in need. However, the results obtained were not too impressive, which is due to the limitation of the design and user compliance. The main concern on the design issues with system portability. Some users expressed dissatisfaction with the poor driving performance, the difficulty of donning and doffing, and discomfort while wearing the orthoses. Also, most of the currently available flexible spinal orthoses are constructed out of strong fabrics or elastic materials with a variety of stiffer supports. The result of its application is disappointing as the clinical effectiveness is low. On the other hand, the proposed multiple-joint spinal orthosis in this research is constructed using multiple segments that are similar to human spine biomechanics. This orthosis is designed for healthy individuals, back pain, and spinal cord injury patients. The proposed multiple-joint spinal orthosis should be able to support and self-control different segments of the three flexible curves of the human spine according to the subject's need. This feature is feasible as the orthosis will be implemented with an integrated mechatronics system that stored the active-passive spinal segments library from the fundamental study that has been described earlier. It is crucial to design and

develop a multiple-joint spinal orthosis that will give a high impact on rehabilitation, muscle training, and human physical support.

1.3 Research Questions

- (a) Which analysis approach is more suitable to study the thoracic and lumbar spine during critical daily activities when comparing between the single and multi-segment spinal?
- (b) Which muscle activation contributes to active and passive spinal segments in critical daily activities?
- (c) How to determine active and passive spinal segments based on the motion capture analysis?
- (d) What are the correct endpoints for the thoracic and lumbar regions in order to get an accurate study of spine kinematics?
- (e) Is there any contribution from the hip and knee angles with the spinal kinematics in critical daily activities?

1.4 Research objectives

The main objectives of the research are as follows:

- (a) To analyze the postural movement during spinal motion in critical daily activities.
- (b) To determine the active and passive spinal segments of thoracic, and lumbar regions and their contribution to spinal motions during critical daily activities.
- (c) To develop a multiple-joint spinal orthosis for assisting the spine of healthy individuals to supports the thoracic, and lumbar spine.
- (d) To implement an integrated mechatronics system on the developed multiple-joint spinal orthosis.

1.5 Scopes

The scopes of the study are as follows:

- (a) The study is focused on the dynamic daily tasks of asymptomatic individuals. The study focuses on Sit-to-Stand, Forward bending, Pick an object on the floor, and Weightlifting of 5 and 10kg loads. Excluding STS, other tasks are then divided into three more tasks that focused on the difference in postural implementation. All tasks are studied for the Begin and Return cycle as well.
- (b) The motion capture and postural detection analysis utilized MAC3D and Vicon. The image processing and data analysis is completed using MATLAB.
- (c) The human spinal motion consists of 6 degrees of freedom (DOF), which are flexion, extension, lateral bending to the left and right, and rotation to the left and right.

However, considering all motions for the design will result in the high complexity of the proposed multiple-joint spinal orthosis. Thus, the flexion and extension motions are included in the design as the literature also agreed that there is little or no correlation of back pain with lateral flexion and rotation.

The research focusing on the development of the multiple-joint spinal orthosis for the cervical, thoracic, and lumbar multi segments that have these characteristics: simple locking mechanism, effectively wearable, excellent stress endurance, and lightweight.

(d) Finite element software, Abaqus 6.14-2 was utilized to simulate the 3D model of the proposed spinal orthosis components to assess the mechanical stress, deformation, and failure process respectively. The spinal orthosis' control system is built using MATLAB software and its toolbox.

(e) The integrated mechatronics of the developed multiple-joint spinal orthosis implemented a PC-based control system.

1.6 Significance of Research (Novelty)

To date, currently, there is no intelligent multiple-joint spinal orthosis that emphasizes the design of the human spine biomechanics, and then provides assistive support and control on the human spine covering from cervical to lumbar that are made up of multi-segments. Most of the studies and related article reviews of the last 10 years only focused on up to two joint or segment mechanisms to support patients with lower back pain. However, the acceptance of the patients was not too impressive due to the limitation of the design and user compliance. Also, most of the currently available multiple-joint spinal orthoses are constructed out of strong fabrics or elastic materials with a variety of stiffer supports. On the other hand, the proposed spinal orthosis in this research is designed with multi-joints segments similar to human spine biomechanics. This orthosis is designed for healthy individuals, patients that suffer from back pain, and spinal cord injury by implementing proper postural, monitoring, and controlling of the spinal curves. The proposed multiple-joint spinal orthosis should

be able to support and self-control different segments of the three flexible curves of the human spine according to the subject's need. Similar studies on spinal kinematics especially for segmental approaches are still few in numbers. Moreover, none of the literature has highlighted the active and passive spinal segments in critical daily activities as they only defined it in terms of high or low mobility. This research also focuses on a wider range of tasks that include both begin and return cycles to develop the data profile of active and passive spinal segments. This data profile will be able to contribute to the rehabilitation purpose significantly. Importantly, this data profile is implemented to the proposed multi-joint spinal orthosis which can give dynamic functionality and enhance the capability of the orthosis. Additionally, the research implements a new palpation method that uses the olecranon and styloid processes of Ulna as landmarks.

CHAPTER 2

LITERATURE REVIEW

2.1 Introduction

This chapter begins with a detailed review of the low back pain that affects Malaysia and the world in many ways. This chapter also discussed in detail the spinal range of motions (ROMs), the effect of age and gender, load analysis, and palpation method used in exploring spine biomechanics. The third subtopic focuses on the topics related to gait and posture such as active-passive spinal segments, the relationship between spinal curves, and critical daily activities, and also on the change of the spine in different activities. The fourth subtopic explores the wearable devices used to assist people with back pain problems, viewed as a state of the art, which is spinal orthosis. The final subtopic emphasizes the research gap of this study based on the literature review discussed in the previous subtopics.

2.1.1 Low back pain

Low back pain (LBP) is a common painful condition that is encountered both in general and specialist practice. A review and analysis study conducted by Hoy, Bain [7] showed that LBP is a major problem throughout the world and is most prevalent among adolescent females and persons between 40–80 years of age. The prevalence of LBP was highest during middle age, which represents some of the most productive years of a person's working life. This results in a major economic impact for many individuals, families, businesses, and governments. However, in Malaysia, the age range differs as the occurrence of the back issue is most common for people aged less than 40 years old [8, 9]. In Malaysia, the prevalence of back pain was found to be 12%,

and it was rated as the ninth and fifth most common complaint in public and private primary healthcare clinics, respectively, between August to November 2012 [10]. In a semi-rural community survey, the rate of LBP was found to be around 12% while in the population at risk such as the commercial vehicle drivers, the rate was quite high reaching up to 60%. About 11.6% out of a population of 2600 in a semirural area of Malaysia was diagnosed with LBP problems [11]. Moreover, it caused a loss in individuals' workdays and the organization's productivity [12]. A study by Lonik, Kamauzaman [9] found several reasons for back pain issues among ambulance workers in Malaysia that are evident among men, which is related to a smoking habit, and active participation in outdoor hobbies. In sum, the factors of back pain in Malaysia include the usage of the non-ergonomic chair, poor sitting posture implementation, the distinct gender-workplace correlation, history of back trauma [8, 9, 13]. Dealing with people who suffer from LBP requires crucial understanding as it affects the physiology of the patients themselves. Understanding and tackling the issue is still a strenuous challenge even for primary care doctors due to many obstacles. Such obstacles include poor awareness and conflicting views regarding the usefulness of the local clinical practice guideline (CPG) in clinical practice, constraints, and a perceived lack of support from multidisciplinary teams [14]. The leading causes of LBP, worldwide, are overloading and bad posture [15]. Excessive and frequent weight lifting at home or workplace also contributes to the LBP [16] which is related to work intensity (weight lifted per hour). "Forward head" posture or an increased lumbar lordosis is also found to be linked to spinal pain [17]. Some preliminary work carried out in the early 1970s found many problems faced by the LBP patients [18]. Such problems are that it will limit the movement of a person and reduce motion efficiency, especially within the spinal motion. Finding from the literature also indicate that back pain affects spine biomechanics. Table 2.1 lists the many forms of LBP injuries, including their classification, causes, symptoms, and segments. It also revealed that the lower lumbar and back muscles are the most commonly reported segment for the problem. The lumbar spine and upper thorax rotations significantly increased to the dominant side in the LBP people during lateral bending [19]. Viggiani, Gallagher [20] also reported the difference in spinal kinematics within the lumbar between healthy and back pain patients. The reported pain was greatest in the full extension with having greater contributions from the higher level of the lumbar. LBP can be more than just physical as it causes the patients to experience physiological stress.

2.2 Spine Biomechanics

Biomechanics is the application of mechanical concepts to living beings to better understand how all of the bony and soft spinal components work together to maintain spinal stability and functionality [21]. Spine biomechanics is a general term that encompasses the study related to the spine's anatomical structure, mechanical capabilities, spinal diseases, and the relationship between various factors, among other things. The anatomy of the spine is complex. The spine, or backbone, is made up of a column of 33 bones and tissue extending from the skull to the pelvis as shown in Figure 2.1. To be specific, the spinal column has 24 articulating and 9 fused vertebrae. In medical terms, the spine is also known as the vertebrae, and the individual spinal bone is the vertebrae. The vertebrae house and protect a cylinder of nerve tissues known as the spinal cord through a hollow path which is known as the spinal canal. The spinal cord and the brain make up the central nervous system (CNS) that carries a signal which controls the body movement and conveys sensation. Between each one of the vertebrae is an intervertebral disk or band of cartilage serving as a shock absorber. These disks also give the lower back flexibility in human motion. The spine is divided into five regions which are the cervical, thoracic, lumbar, sacrum, and coccyx [22]. The vertebra is chained from the cervical to the coccyx and is divided into five regions as in Table 2.2. Every vertebra is numbered according to the anatomical height for the cervical, thoracic, and lumbar regions, as shown in the table. The vertebra on the top is numbered as the lowest value, which is 1. The following subsection goes into further detail on spine biomechanics, including spinal mobility, range of movements, the effects of age and gender on spinal motion, and load analysis. Also covered are the techniques and palpation methods utilised in spinal studies.

Table 2. 1 Types of lower back pain injury

Types	Classification	Causes	Example	Symptoms	Segments
Discs	Lumbar disc herniation	A pinched nerve (radicular pain)	Spinal stenosis and bone spurs from spinal arthritis	Numbness, weakness, tingling, and pain in the leg. The lower back pain and/or pain in the buttock	Lumbar spine (L4-5) and (L5-S1)
	Degenerative disc disease	Disc pain (axial pain)		Pain and possibly radiating weakness or numbness	Lumbar spine (L4-5) and (L5-S1)
Joints	Degenerative joint disease	Breakdown of the cartilage of the joints and discs in the lower back	Osteoarthritis or spinal stenosis	Cause stiffness or pain in the back, and cause weakness or numbness in the legs if it is severe enough	Lumbar spine (L1-5)
	Compression fracture	Vertebral bone in the spine that has decreased at least 15-20% in height due to fracture	Osteoporosis	Deformity, loss of height, crowding of internal organs, loss of muscle, and aerobic conditioning due to lack of activity/exercise	Thoracic spine (T10-12) and lumbar spine (L1)
	Spondylolisthesis	Slipping of vertebrae forward over the one below		Back or buttock pain, pain that runs from the lower back to the leg, and numbness or weakness in one/both leg	Lumbar spine (L3-4) and (L4-5/most common location)
Muscles	Bruised/contusion muscles	Direct/repeated blows from a blunt object strike part of the body		Cause swelling and pain, the injured muscle may feel weak and stiff	Muscles in the low back
	Muscle strain (pulled back muscle)	Muscle is overstretched or torn		Symptoms may range from a mild ache to sudden debilitating pain	Muscles in the low back
Ligaments	Lumbar sprain	Ligaments are stretched too far or torn		Symptoms include pain that can last for weeks, as well as muscle spasm	Ligaments in the low back
Nerves	Lumbar radiculopathy	Compression, inflammation, and/or injury to a spinal nerve root in the low back		When the pain radiates down the back of the leg to the calf or foot, it would in lay terms be described as sciatica	Lumbar spine (L4-5) and (L5-S1)

Table 2. 2 Details on the regional vertebra of the human spine

Region	Details
Cervical	- The seven vertebrae forming the upper part of the spine. - Labeled as C1 to C7.
Thoracic	- Made up of 12 bones between the neck and the lower back. - Labeled as T1 to T12.
Lumbar	- The five largest and strongest vertebrae and located in the lower back between the chest and hips. - Labeled as L1 to L5.
Sacrum	- Made up of five vertebrae fused.
Coccyx (tailbone)	- Made up of four fused vertebrae.

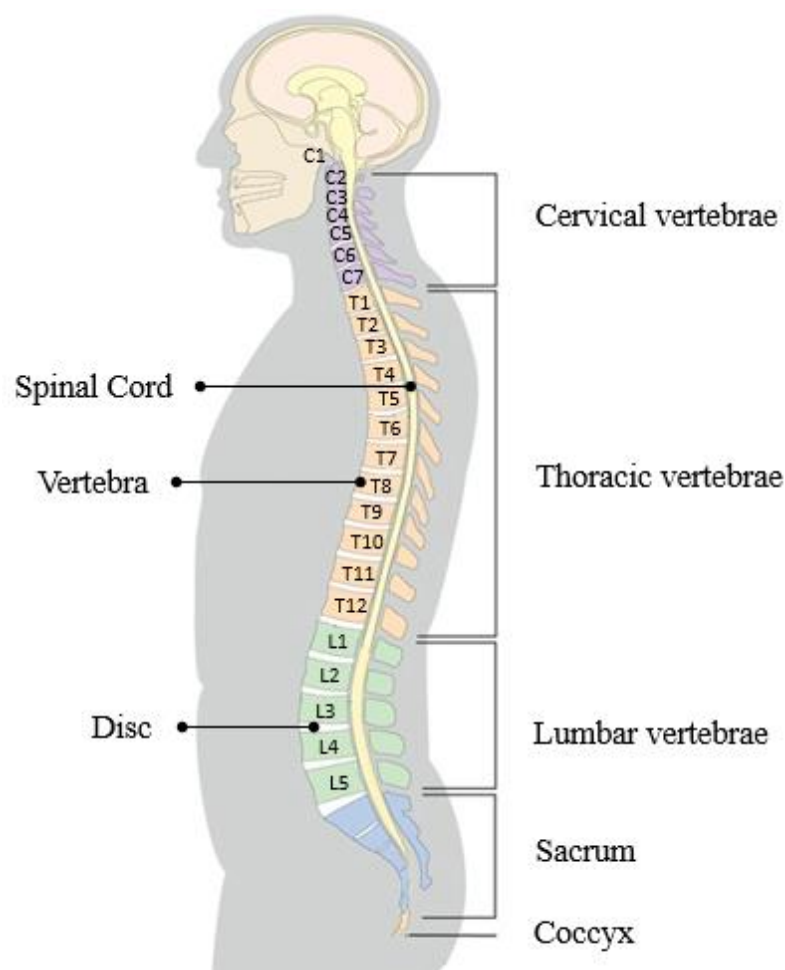


Figure 2. 1 Human spine biomechanics

Source : <http://www.mayfieldclinic.com/pe-anatspine.htm>

2.2.1 Spinal mobility

The human spine has natural curves that form an S-shape. The cervical and lumbar vertebrae curve posteriorly or known as a lordotic curve, while the thoracic vertebrae curve anteriorly or kyphotic curve. Naturally, the spine's curves act as a shock absorber, stability provider, and smoothen the full range of motion throughout the spinal column. Three subsystems that maintain the stability of the spine are the spinal column, the spinal muscles, and neural control [4]. The term 'degrees of freedom' (DOF) is a useful concept used to describe the number of motions of the spine. This number represents a unique independent motion that one vertebra can have to another. The spine has 6 DOF which is made up of 3 translational and 3 rotation motions, along with three axes and each of them have measurable stiffness as displayed in Figure 2.2 [23, 24]. Translational movements are movements of the whole vertebra by the same amount in each direction. There is no change in the orientation of the vertebra. The translation is the "gliding" of the vertebra; it rarely occurs by itself, but often accompanies other movements. The rotational movements of the spine are movements of the vertebra around an axis and all rotations produce a change in the orientation of the vertebra [25]. As illustrated in Figure 2.3, the spine flexes and extends around a sagittal plane axis, bends left and right around a coronal plane axis, and rotates left and right around a transverse plane axis.

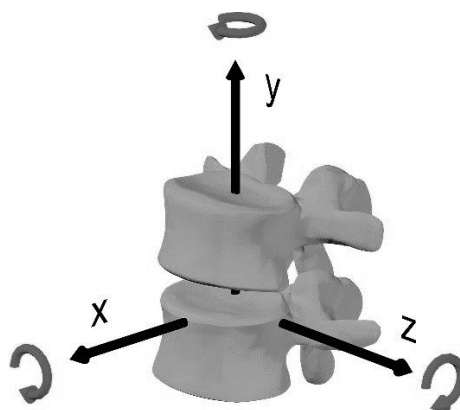


Figure 2. 2 6 Degrees of freedom of a spinal vertebra

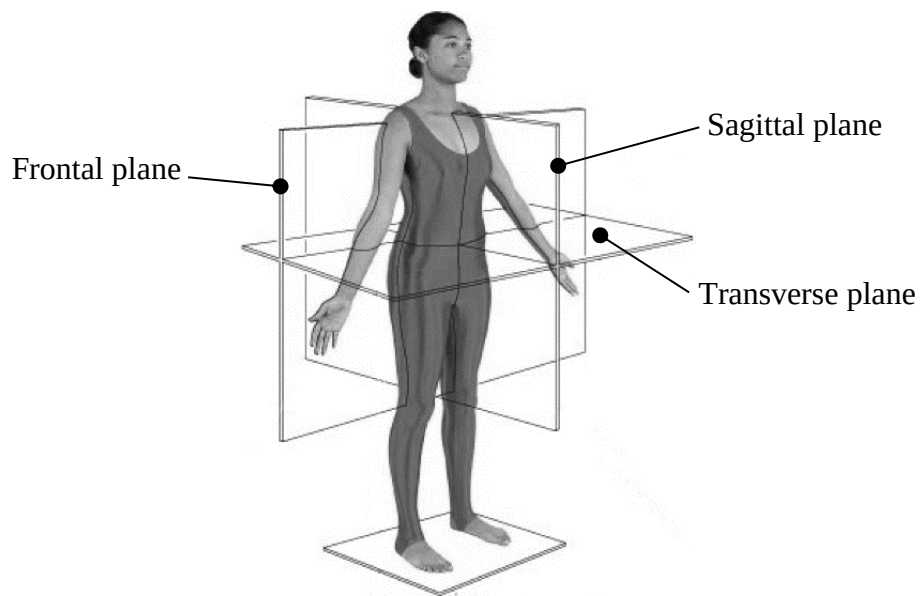


Figure 2. 3 The body planes terminology

Source: <https://www.anatomynote.com/>

2.2.2 Range of motions, ROMs

The flexibility of the spine is limited to certain degrees and the angles differ for each vertebra. Most of the spine movements are only focused on the cervical and lumbar vertebrae and there is only slight movement occurred in thoracic vertebrae. Most of the human vital organs are located within this region and they are protected by a strong cage of bones that is the rib cage and sternum. This structure successfully increases the stability of the thoracic spine by 40% for flexion and extension, 35% for lateral bending, and 31% during axial rotation [26]. Moreover, the rib cage also restricts motion and adds stiffness to the spine [27]. The stability provided by these bones limits the mobility of the thoracic region significantly. Most literature only focused on the cervical and lumbar region as both have greater mobility and are located at the end of the spine.

For cervical, the maximum flexion and extension range up to 80° and 50° , respectively. The normal range for full flexion to full extension is 130° . The maximum range of lateral flexion is 45° . The rotation either to the right or left is up to 80° . For

the lumbar region, the flexes during bend forwards and backward are up to 80° and 30°, respectively. The lateral flexion to the left and right is up to 35°. The rotation either to the right or left is up to 45°. As previously mentioned, the thoracic region ROMs are limited due to the presence of the rib cage and the values are lower compared to the other two regions. Literature studies of the cervical, thoracic, and lumbar ROMs using different non-invasive approaches are shown in Tables 2.3 to 2.6. The literature included in the tables used *in vitro* methods such as the inclinometer and motion capture system, which is safer relative to the *in vivo* technique. Different literature sets the different limits of the spinal region. In the lumbar region, some of the literature labeled T12 or L1 as the upper limit and PSIS, S, or L5 as the lower limit. For the thoracic, most of the literature studies limit the region between T1 to T12. Compared to the thoracic region, the lumbar flexion and extension ranges are substantially wider (Table 2.3 and 2.4). But when compared during lateral flexion and rotation, the thoracic region has greater mobility as in Tables 2.4 and 2.5. Most of the literature on *in vitro* approaches are focused on the range of flexion and extension as these motions are frequently performed in a personalized daily activity [28].

During compression, the cervical tends to have stiffness similar to the thoracic and lumbar regions [29]. In contrast, during other modes of loading, this region tends to have considerably smaller stiffness as compared to the other two regions. The human spine has 6 DOF which are flexion, extension, lateral bending to the left and right, and rotation to the left and right. In completing daily activities or specific tasks, two of the most essential motions used are flexion-extension such as standing from a chair, picking objects from the floor, and reaching for objects.

The endpoint vertebra of the thoracic and lumbar spine is still debatable even though the general assumption of these regions is between T1-12 and L1-5 for both thoracic and lumbar spine, respectively. The spinal curve tends to adjust when a person tries to change his or her posture. Two works of literature assessed the generalization of thoracic kyphosis and lumbar lordotic curve on healthy subjects using two different types of equipment, which are using Magnetic Resonance Imaging [30] and Radiographic [31]. In their critique on this point, Roussouly, Gollogly [31] question the need for understanding the vertebra endpoint critically as the general assumption

for the thoracic and lumbar region may be overly simplistic. Their study found the inflection point of lordosis occurs approximately at T10, not L1 as in Figure 2.4. The characteristics in determining the endpoint of the lumbar lordosis are most dependent on the orientation of the sacral slope and the pelvis. Meakin, Gregory [30] found that the lumbar lordosis curve was found to be intercorrelated between different postures and indicated that despite postural alterations, the lumbar spine has an aspect of form that is partially preserved.

Table 2. 3 Demographic data on the spinal flexion of healthy participants.

No.	Year	Spinal region studied	Instruments	No (Age)		Max Range, in ° (Min, Max)		Position/ Motion	Diff between		Ref		
						Thoracic	Lumbar		Gender	Ages			
1	1994	Lumbar (T12-S2)	Double Inclinometer	1126 (15-65 yo)		N/A	M 54±10 52±9 49±9 45±8	F 63±9 60±10 53±9	Prone lying/ static	No	Yes	[32]	
				M	F								
				122 (16-24yo)	161(15-24yo)								
				295 (25-34yo)	143(25-34yo)								
				159 (35-44yo)	136(35-65yo)								
110 (45-65yo)													
2	2004	Lumbar (L1-5)	Electro-goniometer & a video fluoroscopy unit	100 (21 up to >65yo) M & F: 30(21-30yo), 26(31-40yo), 21(41-50yo) and, 23(51yo >)		N/A	53.0±10.2	Standing/ Dynamic	No	Yes	[33]		
3	2005	Lumbar (T12-S2)	Modified CA6000 Spine Motion Analyzer	405 (16-90 yo) M 209 F 196		N/A	(40,73)	Standing/ Dynamic	Yes	Yes	[34]		
4	2005	Lumbar (T12-PSIS)	Geometric CAD of photographs using reflective markers	128 M (19-59yo)		N/A	Non-shearer 36.1±6.5 (18,49)	Shearer 35.6±6.7 (24,55)	Standing/ Static	N/A	Yes	[35]	
5	2008	Thoracic and Lumbar	Revised skin marker model using video	46 (17-27 & 60-83 yo) 24 young 22 old (15F, 9M) (14F, 8M)		Sit 33.3±8.1 ^Y 40.2±11.2 ^O	Stand 38.3±8.2 ^Y 42.8±11.3 ^O	Sit -15.3±8.3 ^Y 6.0±9.6 ^O	Stand -16.0±5.6 ^Y -15.2±9.3 ^O	Sitting & Standing	N/A	Yes	[17, 36]
6	2016	Thoracic (T1-12) and Lumbar (T12-S1)	Inertial & magnetic sensors	40 M (20-26 yo)		20.5±6.5	50.2±7.0		Standing/ Dynamic	N/A	N/A	[37]	
7	2017	Thoracic (C7-L1) and lumbar (L1-S)	VICON MX	42 (20-80 yo) 8M (20-25yo) 13F (20-25yo) 8M (65-80yo) 13F (65-80yo)		N/A	Young 48.9±8.8	Elderly 37.8±9.0	Standing/ Dynamic	No	Yes	[38]	
8	2018	Lumbar	Epionics SPINE	320 (20-75 yo) 141 M 179 F		N/A	52±10.2 (24.7,76.8)		Standing/ Dynamic	N/A	N/A	[39]	
9	2018	Thoracic, T1-5(Upper) & T5-12(Lower)	Inertial tracking device	22 M (24–28 yo)		19.9±6.4	55.4±12.4		Standing/ Dynamic	N/A	N/A	[40]	

Table 2. 4 Demographic data on the spinal extension of healthy participants.

No.	Year	Spinal region studied	Instruments	No (Age)		Max Range, in ° (Min,Max)		Position / Motion	Diff between		Ref.	
						Thoracic	Lumbar		Gender	Ages		
1	1994	Lumbar	Double Inclinometer measurement	1126 (15-65 yo)		N/A	M	F	Prone lying/ static	No	Yes	[32]
				M	F							
				122(16-24yo)	161(15-24yo)		33±9	26±9				
				295 (25-34yo)	143(25-34yo)		31±8	24±8				
				159 (35-44yo)	136(35-65yo)		28±8	22±8				
				110 (45-65yo)		26±8						
2	2004	Lumbar	Electro-goniometer & a video fluoroscopy unit	100 (21- >65yo) M & F: 30(21-30yo), 26(31-40yo), 21(41-50yo) and, 23(51yo >)		N/A	23.4±8.3		Standing/ Dynamic	No	Yes	[33]
3	2005	Lumbar (T12-S2)	Modified CA6000 Spine Motion Analyzer	405 (16-90 yo) M 209 F 196		N/A	(6,29)		Standing/ Dynamic	Yes	Yes	[34]
4	2005	Lumbar (T12-PSIS)	Geometric CAD of photographs using reflective markers	128 M (19-59yo)		N/A	Non-shearer 14.1±6.8 (2,32)	Shearer 9.8±6.3 (-11,23)	Standing/ Static	N/A	Yes	[35]
5	2018	Thoracic (T1-12: T1-5(Upper) & T5-12(Lower)	Inertial tracking device	22M (24–28 yo)		11.7 ± 3.4°	23.4 ± 10.1°		Standing/ Dynamic	N/A	N/A	[40]

Table 2. 5 Demographic data on the spinal lateral bending of healthy participants.

No.	Year	Spinal region studied	Instruments	No (Age)		Max Range, in ° (Min, Max)				Position/ Motion	Difference between		Ref.
						Thoracic		Lumbar			Gender	Ages	
1	2005	Lumbar (T12-S2)	Modified CA6000 Spine Motion Analyzer	405 (16-90 yo)		N/A		(14,28)		Standing/ Dynamic	Yes	Yes	[34]
				M 209	F 196								
2	2016	Thoracic: T1-6(Upper) & T7-12(Lower), Lumbar (T12-S2)	Digital cameras and a 3D optical motion capture system	44		Only for UT		No LBP		Standing/ Dynamic	N/A	N/A	[19]
				No LBP* (39.7±18.7) 6F,18M	LBP* (43.1±17.4) 8F,12M	No LBP 8.81±1.07	LBP 13.97±1.91	6.20±1.02	9.29±1.06				
3	2018	Thoracic: T1-5(Upper) & T5-12(Lower)	Inertial tracking device	22M (24–28 yo)		24.5±7.4 ^{LD} / 26.5±6.1 RD		16.4±7.2 ^{LD} / 18.3±5.7 RD		Standing/ Dynamic	N/A	N/A	[40]

Table 2. 6 Demographic data on the spinal rotation of healthy participants.

No.	Year	Spinal region studied	Instruments	No (Age)		Max Range, in ° (Min, Max)		Position / Motion	Difference between		Ref.
						Thoracic	Lumbar		Gender	Ages	
1	2005	Lumbar (T12-S2)	Modified CA6000 Spine Motion Analyzer	405 (16-90 yo) M 209 F 196		N/A	(7,7)	Standing/ Dynamic	Yes	Yes	[34]
2	2018	Thoracic: T1-5(Upper) & T5-12(Lower)	Inertial tracking device	22M (24–28 yo)		33.5±10.0 ^L / 37.1±11.7 ^R	7.5±4.5/ 9.2±7.3	Standing/ Dynamic	N/A	N/A	[40]

Abbreviations used: yo: years old, ^Y: Young subjects, ^O: Old subjects, N/A: Not available,

^{LD}: Left direction, RD: Right direction

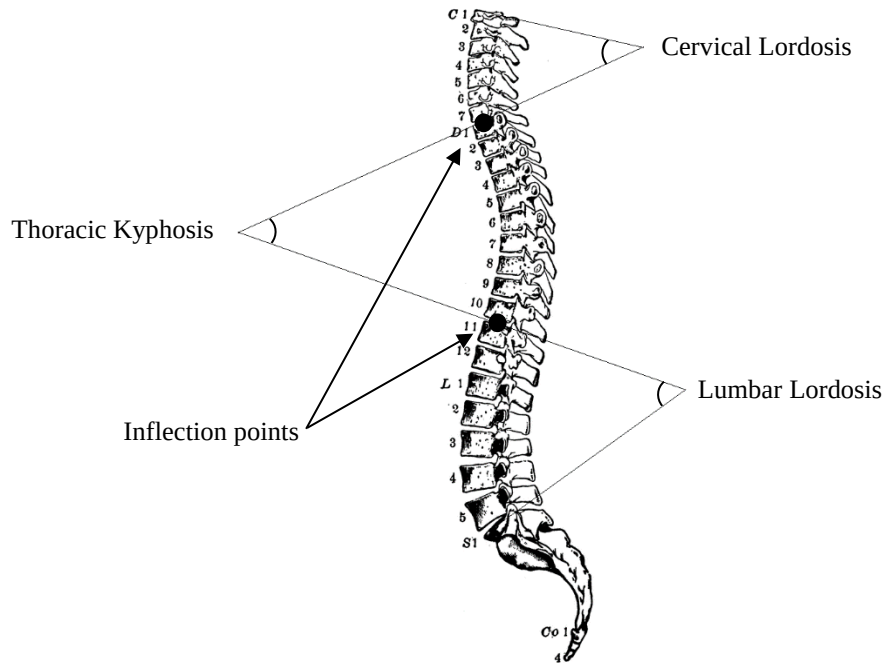


Figure 2. 4 Inflection point of lordosis curve at T10

2.2.3 Age and gender effect on spinal motion

As age increases, a person's ability tends to deteriorate in many ways. Such factors that lead to this include muscle fatigue, in which muscles lose the ability for contraction and relaxation, weaker bones that are prone to osteoporosis, and a reduction in neurological ability. Works of literature are strongly agreed that spinal motion has a strong correlation with age. For the cervical spine, people aged between 70-90 years old displayed approximately 12% less flexion, 32% less extension, 22% less lateral flexion, and 25% less rotation compared with young people aged between 20-30 years old [41]. The high disparity of the cervical spine is also observed within older people. A study by Ignasiak, Rüeger [38] found the difference between regional and segmental displacements and velocities between older and younger subjects within the thoracic region. Furthermore, delay kinematics was observed within this region during standing and sitting in younger subjects (27 ± 3.96 years old) compared with older subjects (70.1 ± 3.85 years old). For the lumbar spine, the increase of age significantly decreases the flexibility of individual motion [32, 35, 42]. A systematic review by Intolo, Milosavljevic [43] found age-related reductions in flexion, extension,

and lateral flexion particularly for 40 to 50 years and after 60 years of age. They also found a minimal age effect on lumbar rotation. Troke, Moore [34] also agreed that the difference in age does not affect lumbar axial rotation.

More recent evidence reveals people older than 50 years have a poor lumbar contribution to spinal movement relative to younger people. As age increases, the deteriorating capability and stiffening of the muscles appear to lead to this critical age-spine correlation. Therefore, it can be concluded that at the beginning of 50 years old, age starts to affect the spinal mobility of a person. It affects the three spinal curves by reducing the ROMs of flexion, extension, and lateral flexion, except for axial rotation.

Literatures studies provide insightful findings according to gender. Young (20-30 years old) and older (70-90 years old) women had greater cervical ROM compared with men [41]. For the thoracic spine, minimal to no gender differences were found. [38, 44-46]. Similarly, the study by gender correlation is not significant within the lumbar spine as it does not affect the spinal motion [33]. More recent evidence [47] reveals that no differences in the maximum lumbar angle during seated or standing forward bending were found between the sexes. Some preliminary work carried out in the early 1990s using a large number of subjects found no significant effect of gender on spinal motion. These studies were conducted by Dvořák, Vajda [42] and Sullivan, Dickinson [32] to assess the ROM of the lumbar spine in 104 and 1,126 healthy subjects, respectively. The lumbar's contribution to trunk motion also had been studied and it was found to be smaller in women compared with men, and at a faster pace [44].

2.2.4 Analysis of loads

In understanding the complex kinetics and kinematics of the human spine, two types of approaches normally used are either *in vivo* or *in vitro* for the scientific study. The *in vivo* method is conducted inside the human body while *in vitro* occurs on the outside. Although the *in vitro* approaches can provide a controlled environment for an experiment, the results must be considered carefully as the accuracy and reliability can be questioned as the contribution of various body parts may be overlooked or ignored.

Additionally, the *in vitro* method offers the safest way of understanding the spine that does not require complex medical protocol and authorization from medical experts. The modeling technique is another method that has been used but comes with different variables, gains, and losses.

Two types of forces that are common and need to be sustained by the spine are the compressive and shear forces, which depend on different tasks. As stated by Schmidler [22], the lumbar vertebrae are the largest of the vertebrae because of their weight-bearing function supporting the torso and head. Each of the lumbar vertebrae supports a different magnitude of load due to different motions and activities. The difference in the structure of each region of the vertebrae explains the magnitude of load that they can sustain. Also, the vertebrae in this region have a higher height compared to the other two regions [48]. As concluded by Hutton, Cyron [49], the shear forces sustain by the lumbar region are very small compared with compressive force. They also concluded that the compressive braking load that the lumbar region can sustain varied between 810 N to 15559 N. They also found that the lowest lumbar, L5, supports the greatest load compared with other lumbar segments and have a high probability of fracture. Moreover, the differences in compressive strength between the upper and lower lumbar vertebrae were not significant. It is apparent from Fig. 2.5 the anatomical dimension of the vertebra from cervical to lumbar tends to rise, which corresponds to the increase in compression strength. It is vital to remember that the cervical and thoracic spines are meant for movement and flexibility rather than load-bearing like the lumbar spine.

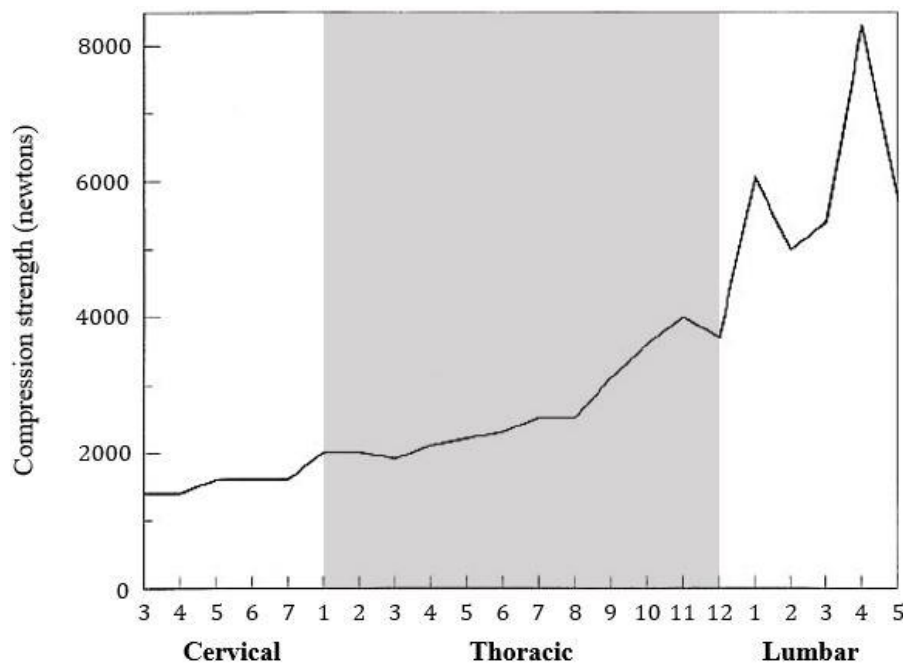


Figure 2. 5 Compression strength of spine

Source: *Benzel EC: Biomechanics of spine stabilization, New York, 2001, Thieme, p1*

2.2.5 Approaches in spinal kinematics

One of the most common assessments conducted for proper treatment and to guide the diagnosis of musculoskeletal conditions and LBP is the postural assessment. Two curves that are commonly assessed during this study are thoracic kyphosis and lumbar lordosis. It has been suggested that deviations in one or both variables may increase a person's risk of developing LBP [17, 50-53], reducing cervical mobility [54], and increasing shoulder range of motion (ROM) [55]. The change of these curves is affected due to several reasons such as the morphological adaptation of both the vertebral bodies and intervertebral discs [56], the use of high-heels during walking [57], and increase of age [58]. Most notably, a recent study by Chokshi, Kadom [59] found a high correlation between lumbar lordosis and Congenital Lumbar Spine Stenosis (CLSS). As stated by Van Blommestein, MaCrae [60], standing radiographs are the gold standard method for measuring spinal angles. The radiograph allows the implementation of the traditional Cobb method, modified Cobb method, computer-assisted method to determine the radius of thoracic spine curvature, and thoracic vertebral centroid angles [61-63] and they produce high-reliability output for medical purposes [64]. However, stricter medical needs were required for the use of

radiographs on human subjects. Besides, the risk of exposure has led to reduce dependency toward this approach even though high percentage of medical institutions still use it. Non-invasive methods (Computed Tomographic, Photogrammetric, MRI, Digital programmed Inclometers, Epionics spine, sensors, etc) are likely to be implemented nowadays as they also produce high reliability results [57, 60, 61, 65, 66], and only small percentage of this approach result in error [67]. Some literature also tried to explore the difference and reliability of invasive and non-invasive approaches simultaneously in understanding the spinal kinematics during daily tasks. Results obtained using electrogoniometric and radiographic methods were slightly different but reasonably well correlated [68].

Many researchers try to study the kinematic of human motion, especially the spine using different approaches (including plain X-ray films, MRI, sensor, etc). Radiographic studies are often accompanied by the same issue, which is the concern of exposure of the subjects and complicated health ethics which require special procedure [69] even though this method was declared as the gold standard of angle studies of the spine. To fix this, other researchers switch to non-invasive approaches, such as MRI [70], garment prototype using strain sensors [71], Wearable Smart T-shirt [72], photogrammetric approach [73], inclinometers [60, 67, 74], electromagnetic tracking [75] and many more. Implementation of a non-invasive study (marker-less and marker-based motion capture system) on the human spine is preferred by many researchers as there is no significant difference between both approaches as they produce almost similar results [76]. The main problem encountered while implementing marker-based analysis is the validating issue as there is still a lack of evidence and findings on the related study [77]. Also, this approach has been proven and effective in predicting segmental cervical and lumbar vertebral angles [78-81].

To explore spinal kinematics, it is crucially important to use a reliable and highly accurate motion capture system that can measure even the minimal changes in the spinal segments. KINECT (as in Figure 2.6 (a)). and 3D Investigator are a few of the systems that have been utilized by literature to study and record human motion. However, the accuracy and reliability of the results obtained from both systems cannot be used as the error is relatively large. Although KINECT has been used for spinal scoliosis assessment[82] and measurement of sagittal spine curvature[83, 84], this

method cannot be used for the segmental study. Moreover, most of these studies only focus on using the depth-sensing function, not the skeleton tracking features. KINECT's Skeletal Tracking System can track up to 6 people and tracks 25 joints per person. However, this feature can only assess 3 large segments of the spine as in Figure 2.6(b), thus questioning the reliability of this system to measure the change of spinal segments which is relatively small. Also, the KINECT device has a low and variable sampling frequency (25–35Hz), which is unchangeable [85, 86]. This will be very problematic in quantifying the position of the body or the joint angle at individual key events, particularly when movement is fast.

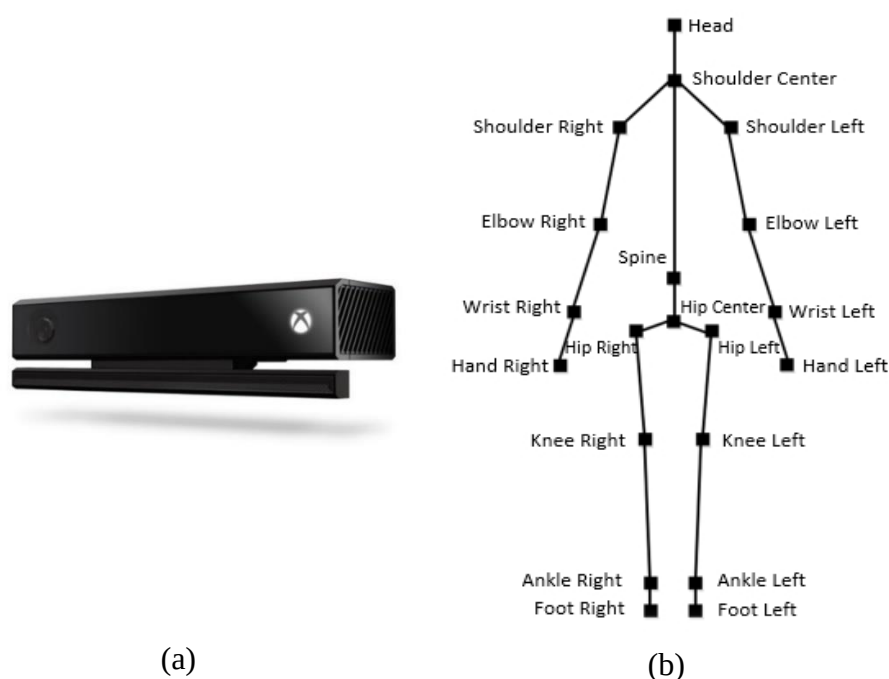


Figure 2. 6 Skeleton tracking with KINECT SDK (Source: Jana [87])

Similar to KINECT, system reliability issues also arise with the use of a 3D Investigator (Figure 2.7). This system has an issue regarding the placement of the active marker on the subject's body as each marker is connected to a single receiver that can only receive a minimum number of marker connections. Besides, the jacket of the wires is designed with poor flexible material, which makes it more difficult to place within the back of the spine as in Figure 2.8 (a) and (b). The main drawback of using this motion capture system is during operation, each sensor must correctly be aligned with the receiver cameras; otherwise, the connection between active markers

and receiver will be disrupted, thus making it more difficult to record even simple spinal gestures such as forward bending and rotational movement.

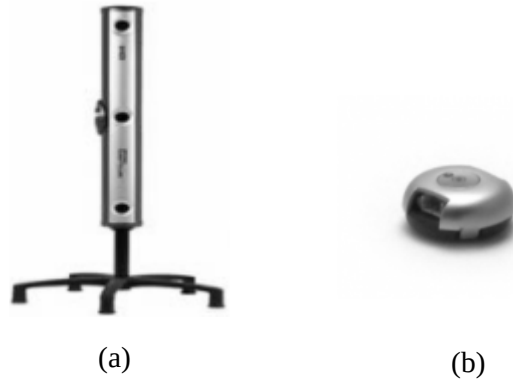


Figure 2. 7 3D Investigator (a) Receiver and (b) Active marker sensor

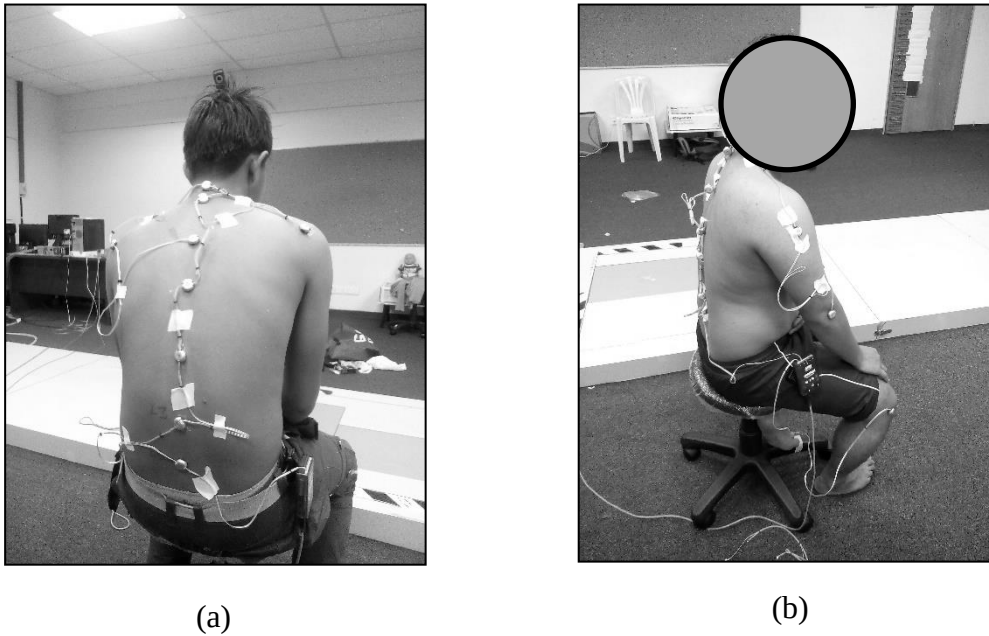


Figure 2. 8 3D Investigator sensors' placement on the subjects' body

Inconsistency in the previous studies may result from the variations in the measurement method used to study cervical mobility and spinal kinematics during sit-to-stand (STS). Research has tended to focus on the lower spinal region (i.e. lower thoracic and lumbar spine) rather than highlighting the contribution of the cervical region. An additional problem is there is still limited study on the dynamic kinematic and assessment focusing on the kinematics of the cervical and thoracic spine during this task. Most of the time, the cervical region is taken out for the assessment of the spinal region during STS. Sforza, Grassi [58] studied the cervical range of motion by

immobilizing the trunk with the application of a stick positioned between the elbows thus limiting the thoracic regional contribution. A previous study that considered the adjacent region of the cervical, which is the thoracic spine, also restricts the whole spine contribution and movement by applying an inelastic strap across the pelvis to minimize the lumbosacral region [88]. There is still considerable disagreement concerning these approaches as it limits the actual kinematics of the cervical and thoracic in active spinal movements during common daily tasks.

2.2.6 Palpation methods of the spine

In most *in vitro* studies of the spine, the palpation technique is the most used method as it provides the safest, quickest, and convenient approach compared with other methods that required more sophisticated and complex instrumentation. This technique is very important especially during spinal treatment and diagnosis of the critical and injured spinal region. One of the main concerns that arise by implementing this approach is the accuracy level. Literature agreed that a high percentage of accuracy and consistency by using this approach is obtainable as the palpation accuracy is associated with between-therapist variability [38, 89]. Four techniques that are commonly used by the therapist, physiotherapeutic, and medical experts are motion palpation (MP), palpation via the posterior superior iliac spines (PSIS technique), palpation via the iliac crests (IC technique), and motion palpation using the previously visible marked landmarks PSIS and CI (MPb). The study by Merz, Wolf [90] found that the accuracy of palpation was greatly improved by combining two or more palpation techniques as compared with just a single technique used. They claim that the accuracy of the single techniques ranged from 45% to 61%. A combination of two or three techniques that were in an agreement can significantly increase the range of accuracy's level from 58 to 78% and from 69% to 83%, respectively. Nevertheless, using more advanced and reliable medical instruments such as ultrasonic imaging has proven to give more accurate palpation landmarks especially when the need arises to study the lumbar motion [91]. The only drawback that concerns the use of the modern approach is it consumes much time and cost as compared with palpation.

For cervical, two palpation methods commonly used are conventional and flexion-extension/motion. The conventional method of cervical palpation defines the most prominent cervical spinous process as the last cervical, C7. The latter palpation technique requires the identification of the two most prominent cervical spinous processes by placing the investigator's or physiotherapist's index and middle finger with the seated subject's cervical spine in flexion. The subject is then asked to perform the cervical extension movement while the fingers are maintained on the vertebra. If the index finger moved anteriorly while the middle finger remained stationary, the lower cervical spinous process would be labeled C7. However, if both fingers stayed stationary, the top cervical palpation would be defined as C7, and the procedure will be repeated by shifting 1 level cephalad at a time to validate the level of C7. Comparing these two methods, Shin, Yoon [92] conclude that the flexion-extension method is more precise than the conventional method used to classify the cervical vertebral. The percentage of accuracy when using the flexion-extension technique (77.1%) is significantly higher compared with the initial approach (37.5%).

2.3 Gait and Posture

2.3.1 Active-Passive vertebra during critical daily activities

The cervical and lumbar vertebrae curve posteriorly, while the thoracic vertebrae curve anteriorly. When viewed from the side, the spine has a natural S-shaped curve. The spine curves respond differently depending on the motion that has been carried out by a human. Specifically, only some vertebrae, either in the same or different regions, tend to respond to the human motion or movement while the others passively fix at their place during the transition of postures. For example, during weightlifting from a standing posture, the fifth vertebrae tend to have higher mobility as compared with the fourth and other upper lumbar vertebrae [93].

Literature and findings on the active and passive vertebrae during everyday motions are shown in Table 2.8. As shown in the table, the spine's curves that have the highest mobility are at the lumbar curve. Kumar [93] reported that the lower lumbar vertebrae show great mobility, especially when in standing and lifting weight postures. Also, it was found that the lumbar flexion tends to be delayed until the weight has reached half of the height lifted [94]. This height was defined as critical height. During standing and forward bending at which the lumbar region reaches its full extension, the fourth and fifth vertebrae showed the greatest movement as compared to the upper lumbar region [95]. It was also found that the mobility of the lumbar vertebrae tends to increase as it goes down from the first to fifth lumbar vertebrae.

In the thoracic region, the motion is more limited due to many reasons. Panjabi, Brand [96] reported that the flexibility of the thoracic region was more significant during flexion compared with extension. Flexion and extension are more limited in the upper thoracic region. Rotation and lateral flexion to the left and right is more limited in the lower thoracic spine. An extension is limited by the ribs, anterior longitudinal ligament, contact of the spinous processes, articular facets, and disc structure. Rotation is mainly limited by the ribcage (ribs, cartilage, and articulations). The significance of these structural limitations increases with age.

Some preliminary work carried out in the early 1990s agreed that the cervical spine's mobility tends to decrease as age increases [41]. In daily activities, humans only use a minimal function of this region as compared with the lumbar region unless the job scope requires them to rotate their heads frequently. There is still limited research focusing on the motion of this region. As stated by May and Lomas [97], the flexion of the cervical curve was observed only during sitting and forward bending. Besides, the largest intersegmental flexion-extension occurs between C4/C5 and C5/C6 [98]. Anderst, Donaldson III [99] reported that the lower cervical spine seems to contribute more to head motion specifically during the end of the ROM. They also prove that the cervical spine motion contribution for right and lateral bending is mirrored to each other and it is also the same for the left and right rotation.

Table 2. 7 Active and passive vertebrae on human common daily motions based on the spinal region.

Daily posture	Spinal Region		
	Cervical	Thoracic	Lumbar
Normal Sitting, Prolonged sitting (>10 mins)	- Lower cervical spine flexed & more extension at the upper cervical spine in the elderly [17].	- Increased thoracic kyphosis in older adults displays poor posture implementation.	- The lordotic curve was nearly 50% lower on average than standing lumbar lordosis [100]. - Older subjects (60-83 yrs) had less lumbar flexion than younger subjects (17-27 yrs) participants. - Flat and slump posture observed [101]. - The lordotic curve almost similar to standing with females had greater lordosis compared with men [101]. - Slump: subjects maintained 44.4% of full lumbar flexion for 20 min [102].
Stooping and weightlifting	Not yet covered	Flexion [94]	- At the early stage, the lumbar vertebrae remain fixed to each other (lift weight starts with 10kg) [93]. - Lumbar flexion is delayed until the weight reaches the critical height [93] or half of its height [94]. The larger the weight (biomechanical hazard) the longer the delay. - Lower lumbar was found to be more mobile than the upper one. - Flexion occurs until the weight touches the ground.
Sitting	- Older adults had a greater flexion in the lower cervical spine and more extension of their upper cervical spine compared to younger adults [17].	- Increased thoracic flexion in old adults[17].	- Lumbar lordosis was nearly 50% lower on average than standing lumbar lordosis [100].

Table 2. 7 Continue...

Daily posture	Cervical	Spinal Region Thoracic	Lumbar
Sitting and bending	- Flexion of lower cervical vertebrae [97]. - The largest intersegmental flexion-extension occurs at C4/C5 and C5/C6 [98]. - More contributions from the lower region during the end of the ROM [99].	Flexion [97]	- Flexion of lumbar [97]. - During forward bending, lumbar flexion contributed more and the hips and lumbar spine contributed almost equally to middle forward bending. Hips had a more significant contribution to late forward bending [103].
Standing	- Older adults had a greater flexion in the lower cervical spine and more extension of their upper cervical spine compared to younger adults. Increased forward neck slope[17].	- Increased thoracic flexion in old adults[17].	- Increased extension in the lumbar region in old adults[17]. - Extension of the lumbar spine[104]. - Greater lumbar lordosis[100]. - Females had greater lordosis compared with men[101].
Standing and bending forward / flexion	- The lower region tends to contribute more during the end of the Range of Motion (ROM) [99].	- A delayed/sequential motion type in young (27 ± 3.96 years old). - Simultaneous motion pattern in the elderly (70.1 ± 3.85).	- During full extension, the fourth and fifth vertebrae showed the greatest movement. [105]. - Maximum total ranges of lumbar flexion and extension were 91° and 81° respectively [106].
Standing and weightlifting	Not yet covered	Not yet covered	Not yet covered

Table 2. 8 Active-passive vertebrae, muscle during critical daily activities (based on general relation)

Year	Focus of study	Findings	Ref.
2013	Kinematic differences within regions of the lumbar spine during STS.	Both lower lumbar spine (LLS) and upper lumbar spine (ULS) regions made different contributions to STS. There was a significant gender difference between both regions when modeled the lumbar spine as two regions compared to a single region.	[107]
2005	Study on the sagittal movement and their relationship between the thoracic, lumbar spine, and hip joints during STS.	During standing, forward trunk lean prior to buttock lift-off (LO) was accomplished by concurrent lumbar and hip flexion. During flexion, the thoracic region begins to extend and resulting in a LO trunk angle of $45.7^{\circ}(\pm 5.8^{\circ})$. Following LO, the hip and lumbar spine extended, and the thoracic spine flexed with the standing thoracic angle approximating the initial thoracic posture in sitting.	[108]
2002	Study the effect of different standing and sitting postures on trunk muscle activity in a pain-free population	Lumbopelvic stabilizing musculature is activated in maintaining optimally aligned, erect postures, and that these muscles are less active during the adoption of passive postures (slump sitting and sway standing) compared to erect posture which resulting in excessive load at the lower spine and leads to lower back pain.	[109]
1997	Patterns of the lower thoracic and lumbar spinal segments and the pelvis during walking.	Proved that the spinal segments move in response to the motion of the lower limbs. It is suggested that the counter-motion of the trunk and pelvis occurs about a variable locus depending on the direction of movement, primarily in the lumbar spine. This is evidenced concerning forward flexion and lateral flexion.	[110]

2.3.1.1 Sit-to-Stand, STS

STS is one of the primary indicators of a person's ability for independence which involves the action of moving upper body mass in the horizontal and vertical direction [111]. STS is divided into two phases, which are pre-extension and an extension phase [112, 113]. During the pre-extension phase, horizontal linear momentum is automatically generated which is then used to transfer the upper body mass forward, approximately near to the feet. As the mass was placed near the feet, the extension phase was then triggered to move the mass in an upward direction into a standing posture. Asian adults spend approximately 7.6 ± 2.4 hours/day (including transportation) in a sitting posture [114]. Literature has agreed that prolonged sitting did not contribute to a healthier lifestyle as it brings more negative effects to the body such as the greater activity of the longissimus thoracic muscle [115], flattening the lumbar curve, and higher intradiscal pressure [28, 116] during the day, even though sitting causes the least fatigue compared to standing [117]. Claus, Hides [118] analyzed the effect of different sitting postures on the thoracolumbar spine and identified 4 common postures during sitting which are slump, flat (at both regions), long (lordotic at both regions), and short lordosis (thoracic kyphosis and lumbar). Among these four, the latter one was suggested to be the optimized posture in reducing the LBP effect.

2.3.1.2 Forward Bending / Flexion

Flexion movement is the most frequent action performed daily, with approximately 4400 flexions performed [28]. The biomechanics of the forward flexion requires the combination of spinal flexion and pelvic rotation which is also known as lumbopelvic coordination or rhythm [119, 120]. During forward flexion as shown in Figure 2.9, contrary forces affect posterior and anterior structures of the spine. The posterior structure (posterior portion of the disc, posterior ligaments, and muscles) is subjected to compression while the anterior are subjected to tension [121]. A study by Pollintine, Dolan [122] confirmed the increase of anterior intervertebral disc during forward bending. Moreover, the ability to complete the forward flexion task requires

the activation of back muscle (lumbar erector spinae) flexion-relaxation behavior that can affect the changes in lumbar curvature. Chen, Lin [123] point out that greater lumbar curvature was achieved within flexible participants.



Figure 2. 9 Example of forward-flexion posture in daily tasks

Source: <https://www.strongerbyscience.com/lumbar-flexion/>

2.3.1.3 Weightlifting

One of the earliest works is conducted by Davis, Troup [94] using the Chronocyclophotographic method. The subject's ages ranged from 18-39 years. When lifting an object, flexion occurs in the thoracic region, and extension occurs in the lumbar region. They also found that during lifting, the lumbar extension was delayed until the weight reached half of its height from the ground as shown in Figure 2.10. Slight thoracic flexion was observed. Nordin, Greenidge [124] also experimented to understand the spinal configuration for various loads during lifting. The participants were 15 healthy young subjects aged 20 to 38 years old. The experiments were focused on lifting crates with different weights (0,10 and 20kg) using straight-legs and the bent-over-back method of lifting. They found that cervical and thoracic were not affected by the amount and variation in load. However, the mobility of the lumbar spine tends to decrease significantly as the load increases, which suggests the coexisting of the lumbar-external load relationship.

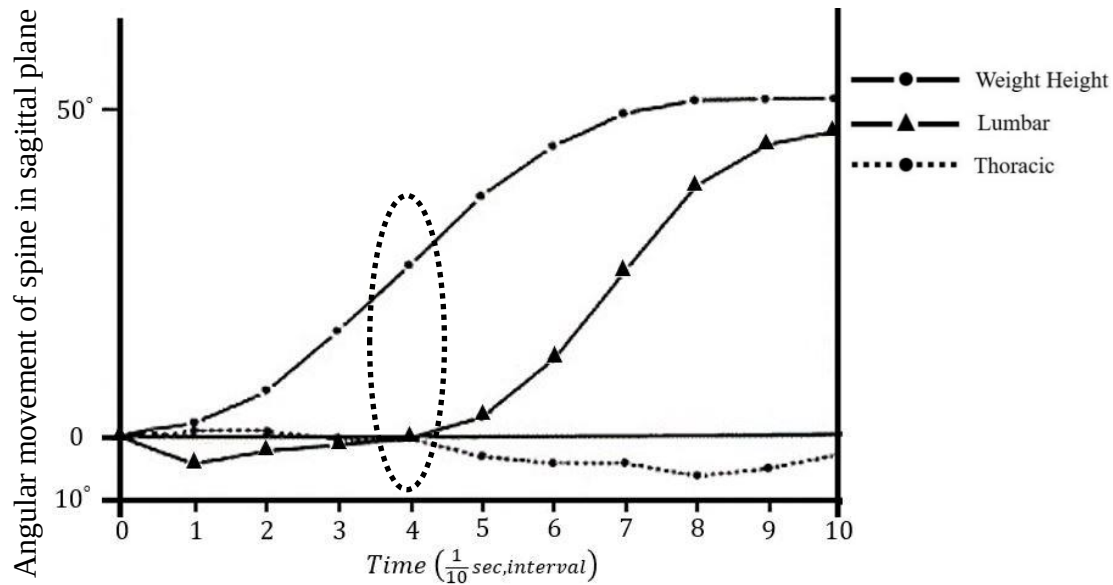


Figure 2. 10 Data for male subject lifting the 20 kg of weight by using the squat method.

Encircled is the point at which the lumbar extension is delayed at the middle weight's height. Source: Davis, Troup [94]

2.3.2 Correlation of spinal regions during critical daily activities

Spinal movements are divided into 6 movements, which are flexion, extension, lateral bending (left and right), and rotation (left and right). Rohlmann, Cons Muller [28] claim that the flexion movement is frequently performed in a personalized daily activity, in which approximately 4400 flexions have been performed. Contrary forces affect posterior and anterior structures during forward flexion. The posterior structure (posterior portion of the disc, posterior ligaments, and muscles) is subjected to compression while the anterior is subjected to tension. Posterior outer annulus fibrosus, zygapophyseal joint capsule, and posterior ligaments play a vital function by providing resistance and limited the bending process [121]. In completing daily tasks, the cervical region used 20-40% of the maximum cervical range of motion [125]. Concurrent cervical and lumbar spine motion with a high number of motions were observed when a person does a bend forward posture (i.e. picking an object up from the floor). Youth displayed greater flexibility of the cervical range of motion (ROM) angle compared with the middle-aged adults in the sagittal and frontal planes [126]. In

general, studies of these critical daily activities only focused on the vertebral and muscle characteristic considering only the lower limb (i.e. hip, knee) and lumbar region [108, 127-135]. Papi, Bull [136] found the spinal segments did not share the same and were predictably kinematic in a specific daily task.

Works of literature reported a high correlation between thoracic kyphosis and lumbar lordosis curves with spinal issues. The dominance of sitting postures during the day [114], in particular in patients and the elderly, leads to a reduction in the lumbar lordosis over a wide range of time [137]. Moreover, advances and availability of present-day technology account for 50% to 60% of the sedentary lifestyle of adults [138], in which most of the activities involve an increase in spinal flexion and poor posture implementation [139]. Korakakis, Giakas [139] also suggested that repeated end-range spinal movement by sitting can abolish the proprioceptive deficit induced by prolonged flexed sitting posture. A study by Harrison, Harrison [140] found that sitting causes the pelvis to rotate backward, reduction in lumbar lordosis, trunk-thigh angle and knee angle, and an increase in muscle effort and disc pressure. The lowest sitting disc pressure and electromyography were recorded in seats with backrests. The use of a backless chair appears to have allowed the adoption of a natural slump posture when seated [101], which then leads to a significant increase in thoracic curvature and decreased lumbar flexion posture. A study by Kuo, Tully [17] found that the large thoracic kyphosis leads to bad posture and contributes more towards the LBP problem and it is more obvious especially in the elderly during sitting and standing posture. Increased curvature in the thoracic spine is associated with higher spinal loads attributable to gravity and muscle force, and a strong linear relationship exists between the magnitude of the load and thoracic kyphosis. Change in the thoracic kyphosis angle is also associated with the changes in the soft tissues and reduction of the mineral content in the bones with the progression as age increases [141]. Also, this leads to an increase in spinal curvature as gradual vertebrae wedging occurs as well as narrowing of the intervertebral disks [142, 143].

2.3.3 Spinal transition and change in height during daily activities

During forward bending maneuvers in manual material handling operations and lifting tasks, while counterbalancing the net reaction moments, the spine is subjected to large compression forces substantially greater than the existing gravity and external loads. The compression force on the lower lumbar segments could in these cases reach and exceed values of 3 to 4 kiloNewton(kN). Occupational tasks with frequent bending and lifting have been recognized as a risk factor in low back disorders. These activities involve large compression loads in addition to large rotations in the forward flexion direction.

Each vertebra between the cervical to thoracic has its allowable range of motion. The cervical vertebra has a greater range of motion in angle compared with thoracic and lumbar vertebrae. The cervical spine is more flexible compared with the other two as it supports the head motion. It has a wider range of motion including rotation to the left and right and flexion either up or down. The lumbar spine also has a wide range of motion but is lower than the cervical. This spine allows the body to be twisted easily either to the left or right. In contrast, the thoracic spine has a minimum range of motion as it is attached to the ribs. This spine is relatively immobile because of the restrictive nature of the rib cage. Figure 2.11 shows the range in the motion of each vertebra for an average person. Based on the figure, the thoracic vertebrae have the lowest range of motion, which is also supported by the study of Liu [144]. Figure 2.12 shows the allowable ROM of the entire spine for Flexion-Extension, Lateral Bending (Left and Right), and Axial Rotation starting from the first cervical down to the last lumbar vertebrae. The cervical spine has a wider range of flexibility compared with the thoracic and lumbar vertebra.

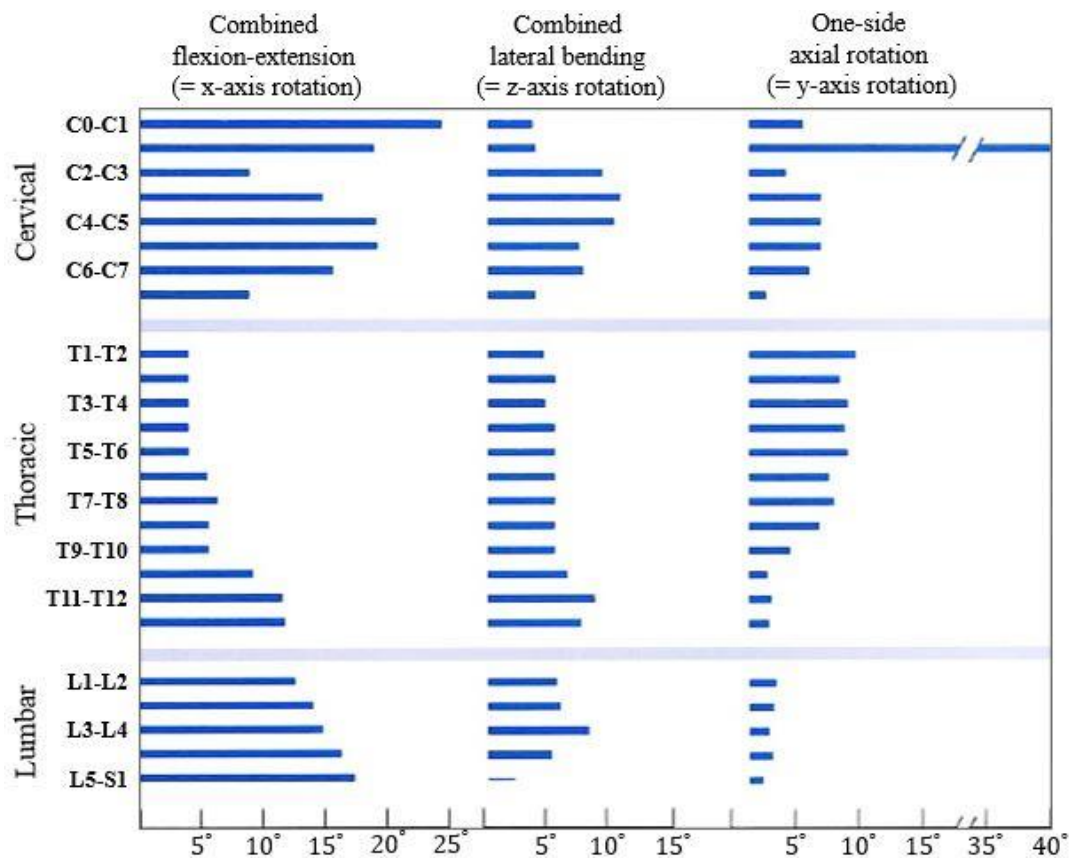


Figure 2.11 Range of motion of the cervical, thoracic and lumbar spine

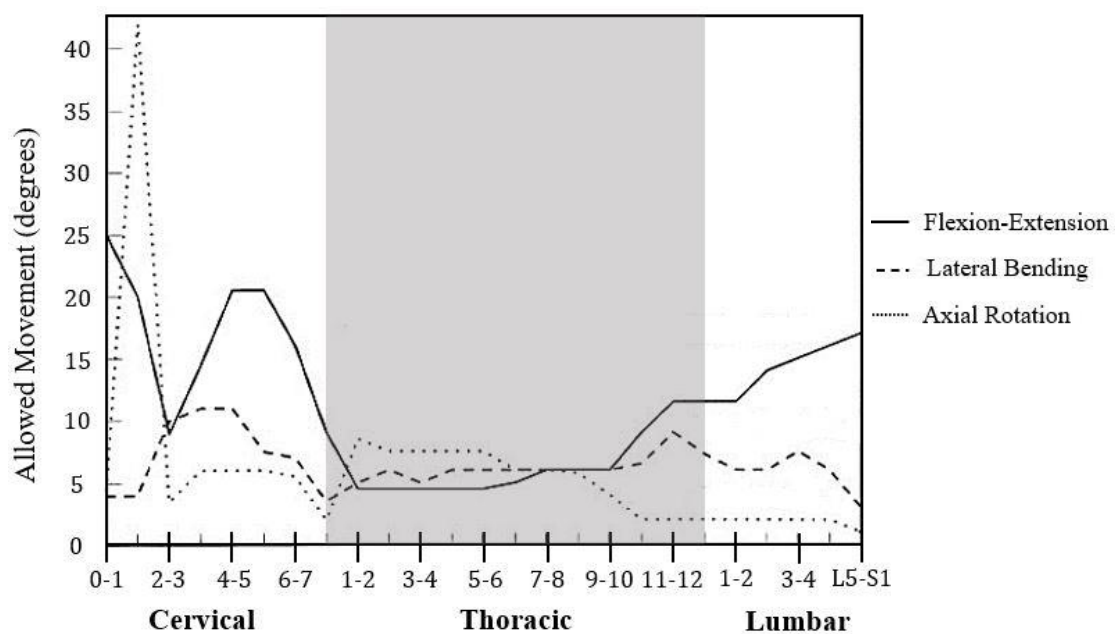


Figure 2.12 ROMs of the spine

Source: Compiled from White AA, Panjabi MM: *The lumbar spine*. In: White AA, Panjabi MM, editors: *Clinical biomechanics of the spine*, 2nd ed., Philadelphia, 1990, Lippincott.

Knowledge of the normal spinal range of motion brings significant benefits in understanding how the various commercial spinal orthoses can limit the range of those suffering from lower back pain. During normal motion and activities, the average height of a person changes frequently but in very small alteration, normally measured in millimeters. The change is due to the compression of the intervertebral disks that affect the average spinal height. Also, the change is significantly observed especially when a person is carrying a load, sitting without the lumbar support, and maintaining a fixed posture for a long period. The reduction of spinal height is also an indicator that the vertebra and intervertebral disks are trying to sustain some amount of load that may lead to LBP for some cases. A study by Poortmans, Brismée [145] using in vivo measurements using a stadiometer found that the change of sitting height was strongly correlated with lumbar spine height changes measured using ultrasonography. There is still a lack of studies related to average height especially in the Asian region and most of the works of literature only focussed on an older range of age (more than 50 years old). Rodríguez, Rodríguez-Calvo [146] proposed a mathematical regression that can output the best regression results, but it is only applicable to the Spanish population. It was also found that the propped slouched sitting with and without lumbar support significantly increased spinal height after a period of trunk loading [147].

2.3.4 Correlation of hip, knee, and trunk movement during spinal motion

The spine reduces the complexity of a normal human being by providing stability in doing most of their work every day, protecting vital organs, and bearing the external load carried by them. Also, it triggered other body parts or *vice versa*. This relation is a perfect match that helps to complete any jobs or tasks demanded by a human. For example, during weightlifting in a standing posture, the trunk movement will activate passive muscle located behind the lumbar spine and assist in the lifting job. The load sustained by the lumbar vertebrae, especially in the lower lumbar region, will reduce significantly. Hip, trunk, knee joint, and other body parts will trigger the spine motion or *vice versa*.

Most researchers and clinicians study human motion especially at the spine region using two different approaches, which are the marker-less and marker-based motion capture systems. A study by Perrott, Pizzari [76] concludes that there is no significant difference between both approaches as they produce almost similar results. The main problem encountered while implementing marker-based analysis is the validating issue as there is still a lack of evidence and findings on the related study (Lundberg, 1996).

During weightlifting, the use of different techniques contributes to a significant effect on spinal motion, especially in the lumbar region. Bending the knees during the lifting effectively reduces the lumbar range of motion and increases the stability of the spine [148]. Studying the lumbar spine in more specific ways such as dividing it into two different regions [107, 127] reveal the significant contribution of the region toward human motion in completing different tasks every day. The lower lumbar spine shows high mobility and velocity for all activities as compared with the upper region.

Methods used during the study and important findings related to spine motion and the human body parts are shown in Table 2.9. It is proven from works of literature that spinal motion affects the knee and lower limb in many ways. During knee flexion contracture (FC), lumbar lordosis, and sacral inclination tend to decrease significantly [149]. Implementing active sitting during prolonged sitting has shown an increase in the trunk motion, resulting in a healthier spine [150]. As a person changes posture from sitting to standing, concurrently the lumbar and hip flexion is accomplished together with forward trunk lean. As the lumbar region flexed, the thoracic region responds by extending until it reaches optimal condition. Following lift-off, the hip and lumbar spine extended and the thoracic spine flexed with the standing thoracic angle approximating the initial thoracic posture in a sitting posture[108].

Table 2. 9 Relation between spinal motion and other human body parts in completing daily activities

Year	Focus of study	Findings	Ref.
2016	Study the spinal movement relative to hip motion by modeling the lumbar spine (single-joint and two separate regions) and evaluate the range of everyday tasks.	The lower lumbar spine (LLS) has a greater velocity and hip ratio compared to the upper lumbar spine (ULS) for all tasks given. Proved that modeling the Whole Lumbar Spine (WLS) underestimated the LLS motion by as much as 37% and over-estimated the ULS motion by as much as 45% compared to separate regions.	[127]
2015	Finding the relationship between the knee joint and spinal alignment and the spinal range of motion	Proved that the knee and the spine affect each other and the spinal range of motion (ROM). The involvement of spinal ROM decreases lumbar lordosis and sacral inclination during knee FCs.	[149]
2014	Effect of prolonged active sitting on the trunk motion.	Active sitting results in increased trunk motion and could have a positive effect on low-back health.	[150]
2005	Study on the sagittal movement and their relationship between the thoracic, lumbar spine, and hip joints during STS.	During standing, forward trunk lean prior to buttock lift-off (LO) was accomplished by concurrent lumbar and hip flexion. During flexion, the thoracic region begins to extend and resulting in a LO trunk angle of $45.7^{\circ}(\pm 5.8^{\circ})$. Following LO, the hip and lumbar spine extended, and the thoracic spine flexed with the standing thoracic angle approximating the initial thoracic posture in sitting.	[108]
2002	Study the relationship between the lumbar spine and hip.	The overall contributions during forward and backward bending were similar, however, the spine had a greater contribution to the early stage of the movement. During lateral bending, it was found to be primarily accomplished by movement of the spine, whereas the hips were the predominant sources of movement for twisting.	[151]

2.4 Spinal Orthosis

2.4.1 Introduction

The orthosis is derived from the Greek *ortho* which means straight, upright, or correct. Different orthoses serve different purposes and body regions covered. Generally prescribed orthoses include Fracture Orthoses, Foot Orthoses (FOs), Ankle Orthoses (AOs), Knee Orthoses (Kos), Ankle-Foot Orthoses (AFOs), Knee-Ankle-Foot Orthoses (KAFOs), and Spinal Orthoses [152].

This thesis only focuses on highlighting the study related to spinal orthosis, thus neglecting the study of other types of orthoses. Spinal orthosis aims to work as prevention and correction of deformities [153], reduction of axial loading, post-surgical stabilization, and biomechanical alignment [154], aiding a weakened muscle, assisting rehabilitation, movement restrictions [155], and increasing independence. The spinal orthosis mechanism successfully prevents and corrects the deformities especially for those who are suffering from Kyphosis by providing external forces by using a balanced pressure mechanism. Dowager's hump, generally known as Kyphosis, is an unnatural curving of the upper back that creates a hunchback appearance in the posture, which is often associated with osteoporosis. There are many causes of Kyphosis such as bad posture, disease or damage, osteoporosis, Scheuermann's disease, Potts disease, and spinal tumors. The study on the effect of spinal orthosis as an aid to those suffering from Kyphosis and similar issues which require the restriction of spinal motions has successfully been proven [156]. "Spinomed" is one of the commercial orthosis products that has successfully proven to improve balance in the elderly with thoracic hyperkyphosis [157]. Daily activities either in-home or workplace force the spine to sustain high compressive and shear forces, especially during the act of lifting. In the lumbar region, the magnitude of the shear forces is small compared with compression forces. The vertebrae are subjected to the greatest compressive forces compared with the cervical and thoracic regions [49]. The spinal orthosis act as a force reducer for the spine as the excess forces that pressed the intervertebral disk by the vertebrae especially in the lumbar region during lifting is significantly reduced. The ideal spinal orthosis takes account of many factors

and aspects such as ergonomic, comfort, functional, fits well, and light in weight. The orthosis must be cosmetically acceptable and easy to use. Due to the high cycle of usage, the design and features of the assist device also take account of maintenance issues, and whether the device can ideally be manufactured locally in case of the device breakdown. All the aspects considered are reasonably different from the 3H's predicting failure which means hot, heavy, and horrible looking. The study by Y. Dionyssiotis [158] showed that most of the spinal orthoses that have been used widely used a three-point pressure system as shown in Figure 2.13. The system implements an action-reaction relationship that applies external forces that have different directions but equal in magnitude by which a single force is opposed by two aligned forces which then resulted to balance forces to the body [154]. The three-point reaction force can be used on mandible, occiput, sternum, thorax, armpit, shoulder, back, ribcage, and pelvis.

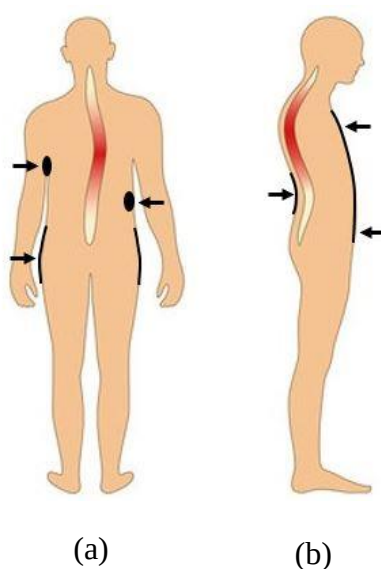


Figure 2. 13 Illustration of the three-point pressure for (a) Scoliosis and (b) Kyphosis treatment.

The study by Seymour [159] found that various factors have increased the use of orthoses such as early fitting, reduction in pain, associated medical problems, psychological counseling, support group interactions, and home and work modifications. Most of the factors obtained relate to the elder group. A flexible orthosis, normally known as a corset, is constructed out of strong fabrics or elastic materials with a variety of stiffer supports. This type of orthosis only restricted less motion and movement as compared with a rigid spinal orthosis. A rigid spinal orthosis

is used when greater control of motion or posture is required. The most rigid orthoses provided the most movement restriction but simultaneously scored worse in comfort. Most of them are fabricated from high-temperature thermoplastics or lightweight metals to reduce the amount of load carried by the user. There are wide varieties of rigid orthoses with a broad selection of pads and coverings.

Generally, the commercial spinal orthosis that has been used widely is divided into three classes, which are flexible, semi-rigid, and rigid. Moreover, most orthosis is designed for a specific function and only aids in a specific region. For example, Cervical Orthosis (CO) is specified to aid the motion of the spine in the cervical region only. Table 2.10 shows the classification of the early design of spinal orthoses that have been used commercially at hospital and rehabilitation facilities based on their focused spinal area to be protected or supported. This table includes rigid, semi-rigid, and flexible orthoses. The restriction of spinal movements is also noted based on the difference of the orthoses in the table.

The design of the passive spinal orthosis has evolved in an intriguing way since 2004, starting with the Spinomed. These modern orthoses have a variety of uses and are aimed at different patient groups, especially for osteoporotic and scoliosis patients. The area of the spine covered by these orthoses differs as well. Furthermore, because of the inadequate feedback of the stiff alternative, the orthoses are designed to be dynamic and flexible rather than rigid. Some of these orthoses offer comparable and distinct movement control mechanisms as a result of various design implementations. Some of these orthoses used rigid plastic rods [160], steel bars[161], and flexible beams[162-164] to prevent excessive trunk and torso maneuver in completing critical daily activities. The Vertebrydyn Strong, Horizon 637 LSO, and SofTec Dorso orthoses use abdominal compression to promote spinal stability, which is beneficial for patients with back pain and rehabilitation purposes. The mechanical architecture of the SpinoCor and V22 Exoskeleton uses trunk control via a harness to transmit upper body load to the lower part, particularly while performing load-bearing tasks. Table 2.11 shows the comparison of commercially available, modern, and passive/unpowered spinal orthosis.

Table 2. 10 Comparison of the early design of the rigid and flexible spinal orthosis

Spinal region	Orthosis	Description and limitation	Control of motion
Cervical	Soft Cervical Collar	Flexion and extension - 5 to 15%, lateral bending - 5 to 10%, and rotation - 10 to 17%	Entire cervical motion
	Miami J Collar	Flexion and extension - 55 to 75%, lateral bending - 60%, and rotation - 70%	Entire cervical motion
	Malibu Collar	Flexion and extension - 55 to 60%, lateral bending - 60%, and rotation - 60%	Entire cervical motion
Cervical and Thoracic	CTO	Provide significantly more restriction of motion compared with CO, spinal is covered from cervical to T5	Flexion and extension
	Halo Device	Flexion and extension - 65 to 70%, lateral bending - 30 to 35%, and rotation - 60 to 65%. Spinal is covered from cervical to T3	Flexion and extension
Thoracic	Jewett Hyperextension Brace	Limit motion from T6 to L1	Flexion
	Taylor Brace	Limit motion of mid to lower thoracic to the upper lumbar region.	Flexion and Extension
	Knight-Taylor TLSO	Cover from top thoracic to medium end of the lumbar region.	Flexion, Extension, and Lateral
	Custom-Molded Body Jacket	Limit motion of mid-thoracic to upper L3 region.	Flexion, Extension, Lateral, and Rotary
Thoracic and Lumbar	Spinomed	The latest design on spinal orthosis, control the movement for all motion within thoracic and lumbar curves.	Flexion, Extension, Lateral, and Rotary
Lumbar and Sacral	Flexible LSO	Corsets or binders.	Entire spine
	Chairback LSO	Limit motion from L1 to L4. Minimal limitation for rotation.	Flexion and Extension
	Williams Flexion LSO	Limit the motion of lateral bending.	Extension and Lateral
	Knight LSO	Limit the motion of lateral bending.	Flexion, Extension, and Lateral
	Custom-molded, plastic LSO	Limit the motion of lateral bending.	Flexion, Extension, and Lateral

Table 2. 11 Comparison of commercially available, modern, and passive/unpowered spinal orthosis.

No.	Orthosis System	Type	Year	Region supported	Target Patients/ Purposes	Control System	Ref.
1	Spinomed	D	2004	Thoracic and lumbar	Osteoporotic patients	- Consists of rigid mouldable support contained in a garment. - Serve as a proprioceptive reminder to extend the thoracic spine.	[165]
2	Thämert Osteomed	D	2008	Thoracic	Osteoporotic patients (45 elderly women)	- Use Velcro tabs to exert pressure in the lumbosacral region - With air chamber pads fixed in the paravertebral & lumbosacral areas.	[166]
3	SpineCor	D	2008	Thoracic and Lumbar	Moderate adolescent idiopathic scoliosis (22 subjects)	- Apply dynamic harnesses on the shoulders, thorax, and pelvic girdles.	[167]
4	Spinomed Active	D	2011	Thoracic and lumbar	Osteoporotic patients	- Consists of rigid mouldable support contained in a garment.	[168]
5	Spinomed III	D	2014	Thoracic and lumbar	Osteoporotic vertebral fractures	- Serve as a proprioceptive reminder to extend the thoracic spine.	[169]
6	Vertebradyn Strong	D	2016	Lumbar	30 healthy volunteers (Driving study)	- Soft belt with integrated dorsal spring steel bars - Apply abdominal compression.	[161]
7	Horizon 637 LSO	SR				- Has a firm back panel and a straightening ventral belt. - Apply abdominal compression.	
8	SofTec Dorso	SR		Thoracic and lumbar		- Carbon-reinforced framework with bars, connected by an elastic fabric - Apply abdominal compression.	
9	Passive Spexor	D	2018	Trunk and hip	Rehabilitation purposes	- Restrict flexion by using a flexible beam.	[162-164, 170]
10	FLx exoskeleton	SR	2019	Trunk flexion	Lifting (Squat)	- Restrict extensive torso flexion and twisting using a rigid plastic rod.	[160]
11	V22 exoskeleton					- Same as FLx. - With shoulder cable to transfer biomechanical loads from the upper body down to the legs.	

*Type: (D): Dynamic/flexible orthosis, (SR): Semi-rigid, (R): Rigid

2.4.2 Multiple joints spinal orthosis

The use of the orthotic intervention in rehabilitation spans the history of humans, from the first crude fracture splint made from sticks in the forest, to the sophisticated modern dynamic orthosis fabricated from hybrid materials. Technology has revamped the field of orthotics, with newer orthoses that are stronger and lighter. Many of the principles have remained the same through time; however, the new materials and structural designs, and breadth of application to a greater number of medical conditions have contributed to the expansive utilization of orthotic intervention. Moreover, there is no evidence that more technologically advanced spinal orthoses provide greater benefits than conventional spinal orthoses [171]. The use of orthotics can be found in almost every aspect of rehabilitation today. With the appreciation and understanding of terminology, materials, generic designs, and the application of orthotics, clinicians can enhance the delivery of healthcare to their clients. The biomechanical principles of orthotic design assist in promoting control, correction, stabilization, or dynamic movement. All orthotic designs are based on three relatively simple principles: (1) pressure, (2) equilibrium, and (3) the lever arm principle. These considerations include and are not limited to the forces at the interface between the orthotic materials and the skin, the DOF of each joint, the number of joint segments, the neuromuscular control of a segment, including strength and tone, the material selected for orthotic fabrication, and the activity level of the client. The following principles provide the foundation for all orthotic designs keeping in mind that the more complicated the orthotic application, the more confounded the various principles become.

The drawbacks of the spinal orthosis cause discomfort, osteopenia (Low BMD), skin breakdown, nerve compression, muscle atrophy with prolonged use, decreased pulmonary capacity, difficulty in donning and doffing the orthosis, difficulty with transfers local pain, psychological and physical dependency, increased segmental motion at the ends of the orthosis, unsightly appearance, and poor patient compliance. Without a good design implementation and strategy, the deficiency from using this spinal orthosis to the spinal patients might not be solved.

From the cervical to the lumbar region, the spine is made up of 24 vertebrae. Thus, to effectively assist the user in completing critical daily activities, the multi-segment aspect should be addressed in the design. An extensive literature review analysis shows that none have tried to control the spinal motion of the vertebrae of the cervical, thoracic, and lumbar curves. The multi-segment anatomy of the human spine is also not taken into account in most current spinal orthosis designs. Most of the multiple-joint spinal orthosis only implemented up to two joints/segments mechanism to support patients with back pain and others were implementing different mechanisms. These joints/segments are differentiate by upper-lower spine[172, 173], thoracic inferior-superior sacrum[174], thoracic-lumbar[175], torso-pelvis[176], and shoulders-upper abdominal-hips[177]. Moreover, these orthoses are simply implemented external flexion and extension forces to help the patients or users in reducing the burden to the lower back during lifting a heavy load with a simple control system. However, wearing this spinal orthosis may encourage a more optimal lifting style in terms of reducing lumbar flexion. These orthoses had implemented different systems, actuators, sensors, and control systems to produce the best spinal orthosis that gives support for those in need as shown in Table 2.12.

Some of the orthoses implemented the spring or elastic belts mechanism to provide external forces for different reasons, which are either to (1) reduce the flexion of erector spinae muscle, (2) control the instability of the lumbar during trunk flexion, and (3) prevent excessive bending, especially during weightlifting task. This mechanism optimizes the stored elastic energy of the spring and thus reducing the dependency on complex and advanced mechanisms. The more advanced orthosis implements pneumatic artificial muscle, linear actuators, and different kinds of motors such as Direct Current (DC), Direct Drive (DD), and Flat motors. Some of these orthoses offer comparable and distinct movement control mechanisms as a result of various actuator implementations. These actuators are used to assist the user by (1) transmitting assistive force and external torque to the body and (2) positioning the trunk in the correct position. Different types of sensors are utilized to provide input to the actuators, depending on the appropriateness of both the movement mechanism and the goal of the orthosis itself. Such sensors used are EMG, bend sensor, gyro and accelerometer, pressure sensor, Force-Sensing Resistor (FSR), Inertia Measurement Unit (IMU), Displacement sensor, and rotary sensor.

2.4.3 Advantages and Disadvantages of spinal orthoses

2.4.3.1 Advantages

From many works of literature and various research studies, it was found that the benefits of spinal orthoses include improving digestive system function, decreasing muscle spasm, improving independent living, improving bowel and bladder function, improving respiratory and cardiovascular system function, decreasing bone osteoporosis, and preventing joint deformity [178]. It has been suggested that orthosis can reduce the average resultant force acting on the lumbar spine up to 9% and 19% using Lumbo TriStep brace and the hyperextension orthosis, respectively. The finding is also more obvious in activity related to sitting than performed while standing [179]. It has been suggested [68] that wearing flexible lumbar belt orthosis has successfully decreased the lumbar flexion-extension movement by 17% and during continuous recording by 22%. Curfs, van Rooij [155] reported that thoracolumbar orthoses influence spinal movement during several daily tasks (including STS) by reducing the bending angle during flexion and extension. Most of the spinal injury patients were recommended to undergo rehabilitation programs such as walking and exercises. Implementing spinal orthoses during daily walking will benefit them further by maintaining good health, decreasing urinary tract infections, improving cardiovascular and digestive systems functions, and improving psychological health.

A recent review by Colobățiu and Mirel [180] on this subject proved that modern spinal orthoses are effective in the management of vertebral fractures in osteoporotic patients. Some works of literature explore the efficiency of three versions of Spinomed for the elderly that are susceptible to osteoporotic disease [165, 168, 169]. The initial version of Spinomed displayed an excellent increment of back extensor strength (72%) compared with Spinomed III (50%) and Active (64%). However, peak reduction of back pain (47%), active-abdominal flexor strength (44%), and limitations of daily living (54%) were more significant with Spinomed Active compared with the other two orthoses. Only Spinomed and Spinomed Active were reported to have successfully controlled the kyphosis angle up to 11% (Figure 2.14). The introductory of air chamber pads fixed in the paravertebral and lumbosacral areas of Thämert

Osteomed resulted in excellent posture correction achievable with 38% of the maximally possible intentional erection [166]. In an investigation into the effect of three different orthoses used in cars, Hofmann, Thumm [161] show that Horizon 637 LSO wearers had better reaction time, brake response time, and brake force compared with wearers of the other two orthoses. The recent design of dynamic spinal orthosis, which is known as Passive Spexor, implemented a flexible beam to limit the user's flexion [162-164, 170]. This orthosis has successfully increased performance in lifting and static forward bending, but decreased in walking, stair climbing and wide stance. Issues related with back pain such as local discomfort was also reduced and those effects were generally bigger for the low-back pain than for the healthy group.

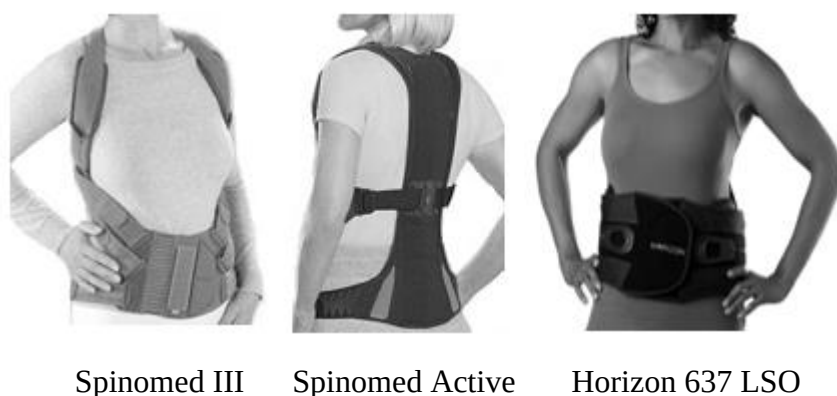


Figure 2. 14 Different designs for flexible spinal orthoses

2.4.3.2 Disadvantages

Even though spinal orthoses are aimed to solve the issues related to the spine, some problems arise together with it. A study by Taghi [178] found that spine orthoses lead to impaired driving performance. Most of the patients that used spinal orthoses due to after surgery or during rehabilitation, encountered poor driving performance through their daily activities. Moreover, Hofmann, Thumm [161] argue that spine orthoses compromise braking performance while in the car. Encouraging patients to perform simple yet repetitive practice during rehabilitation processes is one of the main challenges that are difficult to tackle, especially without the presence of a therapist at home [181]. The evidence that proves the efficiencies of spinal orthoses usage are also few in number and debatable [182]. Prolonged wear of the orthoses required during therapies and care has led to discomfort [155], weakening of the trunk muscles [154],

and poor acceptance by clients and physicians. For example, the Milwaukee brace, also known as Cervical-Thoracic-Lumbar-Sacral Orthosis (CTL SO), which is designed to support the entire spine exteriorly, is needed to be worn for 12 to 18 months, 23 hours a day, with the user, only being out of the brace for exercise or athletic activity. Taghi [178] also concluded that there were many problems reported by patients using the spinal orthoses from various resources such as excessive energy expenditure and mechanical work required, poor cosmetics of the orthoses especially the hip-guided orthosis, a time-consuming problem during donning and doffing (sometimes they required assistance), and experience issues related with fear of falling. Rao, Samant [154] argue that users had a high possibility of getting ulcers due to excessive pressure being applied to the skin. A study by Rohlmann, Zander [179] highlights the variability of spinal load reduction between patients although they use a similar type of orthosis. Such suggestions to make use of the orthoses should be provided together with solid scientific data by the studies [161].

From previous researches, it is found that the conventional rigid spinal orthosis is poorly accepted but still comparable with the flexible one [167]. The rigid spinal orthosis should be worn for a long period of up to years until growth has ceased. Due to poor cosmetic and physical fitness, teenaged patients tend to avoid using it. In contrast, the flexible spinal orthosis comes in a more attractive design and color that looks more appealing in their eyes. It is also reported that the use of rigid spinal orthosis caused the ventilation to be greatly reduced which makes wearing it even less well tolerated especially during humid and hot weather. Although the rigid spinal orthosis was rejected due to its cosmetic and physical appearance, its clinical efficiency was very high as reported by Wong, Cheng [167]. He traces the distinctive failure rate of both types of the orthosis with the flexible type displayed the highest value. Rigid type orthosis displayed a lesser percentage of failure compared with a flexible spinal orthosis. Two new semi-rigid orthoses implement exoskeleton support for the exterior design of the orthosis, namely FLx and V22 exoskeleton. However, Picchiotti, Weston [160] concluded that both orthoses failed to provide a significant biomechanical advantage for loads acting on the tissues within the lumbar spine, joint flexion angle, and moment arms between the L5/S1 joint and the given load.

Table 2. 12 Comparison of existing and newer powered flexible spinal orthosis support for lower back pain.

No.	Orthosis System	Year	Robotic System Types	Actuators and Sensors	Supported movements	Control System	Ref.
1	Wearable Exoskeleton Power Assist System for Lower Back Support	2003	Wearable power assist system	Constant torque DC motor, position, and EMG sensors	Lower back flexion and extension focusing on lifting motion	Speed and position control, and EMG based controller using an artificial neural network (ANN)	[173, 183, 184]
2	Soft Power Suit with Semi-active Assist Mechanism	2008	Lightweight wearable power assist system	DC servo motor, bend sensor, and motion capture system	Waist flexion-extension motion and knee joint	Semi-active assist mechanism using assist force control method	[185]
3	Smart Suit Lite	2008	Passive power assist device system	Elastic belts, force plate	Lumbar muscles (flexion, lateral flexion, and rotation)	Uses motion-based assist method using skin segment and coordinate systems	[185-188]
4	Personal Lift Assistive Device (PLAD)	2009	Assistive device system	A non-motorized system using compression springs, strain gauge, and EMG sensors	Lumbar flexion and extension focusing on lifting and bending tasks	External force generator using the concept of stored elastic energy	[189, 190]
5	Lower extremity exoskeleton	2009	Passive wearable assist system	Spring cable mechanism	Back muscle during squat lifting	External passive extensor moment (restoring moment) generated from stored elastic energy	[191]
6	Power Assist Wear for Low Back Support	2012	Wearable power assist system	Elongation type pneumatic rubber artificial muscles and TPU balloons, gyro and accelerometer sensors	Lower back flexion and extension focusing on lifting motion	Position and pressure control system	[192]
7	Waist Power Assist Suit	2013	Wearable power assist system	Thin direct drive (DD) motors, surface EMG sensors	Waist flexion, extension, and rotation focusing on lifting and twist motions	Simple signal processing, motion recognition based on surface EMG with torque control	[193]

Table 2. 12 Continue

No.	Orthosis System	Year	Robotic System Types	Actuators and Sensors	Supported movements	Control System	Ref.
8	Muscle Suit	2013	Compact and lightweight wearable power assist system	McKibben artificial muscles, near-infrared spectroscopy (NIRS), and pressure sensors	Arm and waist (back support), supported the motion of the upper body	Pressure control system	[194, 195]
9	Active LSO	2016	Actively powered prototype orthosis	A linear actuator, temperature-compensated differential gas pressure sensors	Lumbar flexion-extension (loading & motion)	Position and pressure control system	[174]
10	Passive Spine Exoskeleton	2016	Passive power assist device system	A non-motorized system using extension springs at the thoracic and lumbar spine	Flexion, extension	“Push-pull” external assistive/passive force from springs	[175]
11	CUHK-EXO	2017	Wearable assistive system	DC motor, Encoders & potentiometers, IMUs, Force-Sensing Resistor (FSR) sensors	Lower limb (back, hip, and knee joints) movement during STS and walk	Transmit assistive force from the exoskeleton to the wearer from generated reference trajectories.	[196]
12	AB-Wear suit	2017	Flexible waist assist suit	Straight-fiber-type artificial muscles and a balloon actuator.	Flexion during lifting	The assistive force generated from pressurized actuators	[197]
13	Robo-Mate	2019	Wearable power assist system	Flat motor, Inertia Measurement Unit (IMU) sensor	Flexion, extension, and lateral bending focussing on lifting	External torques based on the change in trunk inclination angle and lifted load	[176, 198]
14	Wearable Power Assist Device (WPAD)	2019	Wearable power assist system	Pneumatic muscles, Displacement sensor	Flexion and extension during weightlifting	Power up the waist by conducting pneumatic muscle contraction	[199]

Table 2. 12 Continue

No.	Orthosis System	Year	Robotic System Types	Actuators and Sensors	Supported movements	Control System	Ref.
15	Active Orthosis (AO)	2019	Assistive device system	DC motors and rotatory sensors	Twisting and lateral bending	Adjust the AO movement based on feedback output controllers given by the manipulators	[177]
16	Active Postural Support Brace	2020	Motion recognition system	Inertial sensors	Trunk motions	Upper-body motion mode recognition using an integrated motion-sensing method	[200, 201]

2.5 Intraclass correlation coefficients (ICC)

The intraclass correlation measures the reliability of evaluations or measurements for data that have been collected or organized into groups. Another related term that researchers used is an inter-rater correlation, which is usually named Pearson correlation. The intra-rater reliability refers to the consistency a single rater has with himself when data is measured across 2 or more trials. In contrast, the inter-rater reliability reflects the difference and agreement between 2 or more raters in the same cases. The intraclass correlation coefficient (ICC) is an index that can reflect both degrees of correlation and agreement between measurements [202]. Two common uses are to compare paired data on the same measure and to determine the accuracy of the judge's ratings of a collection of items[203]. Interestingly, ICC is also related to the concordance correlation coefficient (CCC) and shares the same purposes. The only difference is CCC does not need ANOVA assumptions[204]. ICC values ranged from 0 to 1, with a value closer to 1 indicating excellent reliability. The modern ICC definition is first proposed by Shrout and Fleiss [205] in 1979 with a simpler formula of 6 ICC definitions, compared with originally proposed by Ronald Fisher[206] in 1925. Later on, the formula was redefined by McGraw and Wong [207] in 1996 with 10 ICC definitions that have been used widely in statistical analysis for reliability. Depending on the research, ICC may be reported differently based on the variation of the results. Koo and Li [202] list four questions that can guide the researcher in choosing the correct ICC form, which is, (1) Are similar raters used on all participants? (2) Is the sample of raters randomly selected from a larger population or a specific sample? (3) Does the study concern the reliability of single raters or the mean value of multiple raters? (4) What is the top priority? Either consistency or agreement? Campbell-Kyureghyan, Jorgensen [70]

These questions will guide the researcher to choose the correct and most suitable form of ICC based on the research conducted as in Figure 2.15. Questions 1 and 2 help in the selection of the "Model," Question 3 guides the selection of the "Type" and the last question helps guide the selection of the "Definition". In SPSS, ICC estimation is based on the terminology proposed by McGraw and Wong.

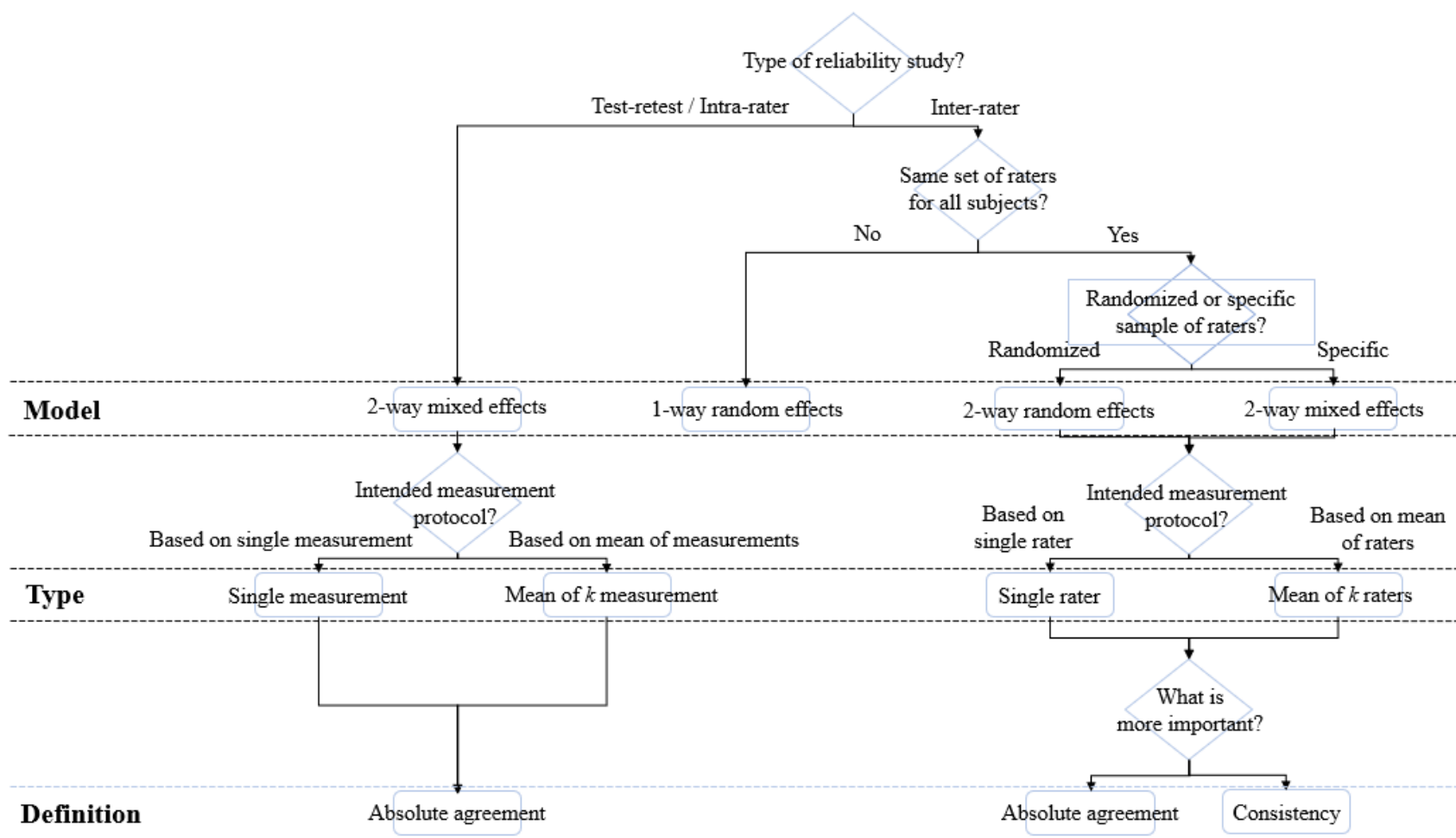


Figure 2. 15 Flowchart of correct ICC selection

Source: Koo and Li [202]

2.6 Research gap

The importance of understanding spinal kinematics in the performance of any activities over a lifetime is rather undeniable. Works of literature agreed that there is a strong correlation between the spine, age, varying tasks, and body components (i.e., hips, knee, trunk, and back muscle) thus highlighting the importance of exploring and understanding the spine critically. However, there are still limited studies that highlight the multi-segmental role of the spinal curve, especially within the thoracic and lumbar curves during frequent and critical daily activities. The definition of the “active-passive” for spinal multi segments contribution in critical daily activities has also not been achieved yet as they only defined it in terms of mobility. This definition is used in this study, specifically in Chapter 4 to differentiate the segments that have a greater or lesser contribution to the tasks studied. Furthermore, the multi-segmental contribution during the Begin and Return cycles has not been dealt with in-depth. Even the data profile of the spinal multi-segment contribution in the critical daily activities has not yet been established. These highlighted issues are of utmost importance, especially in the development of better spinal orthoses that can assist the user or patient optimally. There is currently still no intelligent multiple-joint spinal orthosis that stresses the design of the human spine biomechanics and then provides assistive support and control on the spine, which is made up of multi-segments and spans from cervical to lumbar. In the previous ten years, the majority of studies and relevant article reviews have solely concentrated on one or two joint or segment strategies to help patients with lower back pain. But even so, due to design limitations and user compliance, the patient or user acceptance rate was not very high. Furthermore, the majority of the current multiple-joint spinal orthoses are made of strong fabrics or elastic materials with a variety of stiffer supports. Thus, a more user-friendly spinal orthosis is required for long-term rehabilitation and medical therapy. s

CHAPTER 3

RESEARCH METHODOLOGY

3.1 Description of Methodology

From the literature findings, there is an urgent need to conduct a proper research procedure that emphasizes the difference in spinal segmental contribution in daily tasks. Thus, the methodology of this study is geared toward the aim of analyzing the postural movement and determine the active and passive spinal segments during critical daily activities. To achieve that, this research is divided into three phases. The first phase is focused on literature review and fundamental study on critical daily activity. The second phase is focused on the product design development and control system of the proposed multi-joint spinal orthosis. Using the findings from the literature and conducted studies of the kinematics of spinal segments in the first phase, the integration of the control system, prototype, and the data profile of active-passive spinal segments is conducted in the final phase of this research. The critical daily tasks investigated in this study include:

- 1) Sit to Stand
- 2) Forward Bending from standing posture
 - Straight Hand
 - Straight Back
 - Hanging Hand
- 3) Pick an object on the floor from a standing posture
(effect of the load is neglected as the object's mass is too small)
 - Stoop approach
 - SemiSquat
 - Squat
- 4) Weightlift load of 5 kg
 - Stoop approach
 - SemiSquat
 - Squat

- 5) Weightlift load of 10 kg
 - Stoop approach
 - SemiSquat
 - Squat

The second phase focuses on product design development and control system. For the design development, this study has completed the product design specifications (PDS), morphological chart, selection matrix, and final concept (Appendix A-E). Next, the engineering analysis is conducted to ensure that the design can sustain the specified load when used by the user. The details on the engineering analysis and control system are discussed in Chapter 5 and Chapter 6, respectively. The final phase focus on the integration of the findings of this study (active-passive spinal segments), the prototype, and the control system which is also discussed in Chapter 6.

3.2 Research flowcharts

The operational framework must comply with the stated objectives of this study. The present idea of the operational framework is shown in the flowchart in Figure 3.1 below which is divided into three main phases.

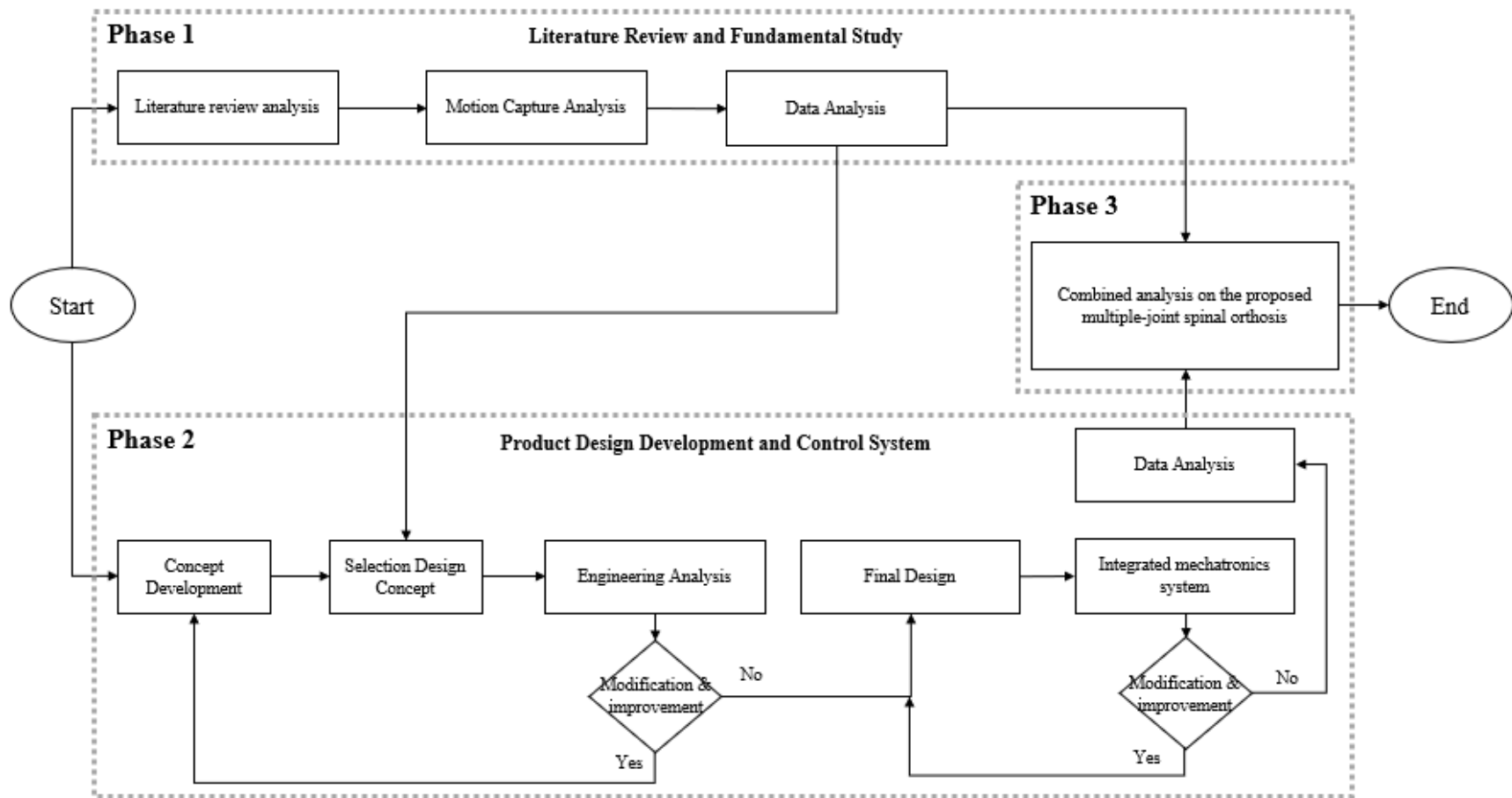


Figure 3. 1 Research Framework

3.3 Motion Capture System

This section will discuss the motion capture systems that have been developed and used through the completion of this research. The implementation of these systems is crucially important to determine the standard posture and active-passive spinal segments during critical daily activities. The findings from this study will be used to design a proposed multiple-joint spinal orthosis. Throughout the completion of the study, a stick marker-based motion capture system has successfully been developed. However, due to a few limitations, the later data collection phase utilizes other motion capture systems such as Kinect, 3D Investigator, Mac3D, and Vicon as these systems have higher accuracy and data reliability compared with the initially developed system. The issues that arise in using some of the systems are also discussed in detail within this section.

3.3.1 MAC3D and VICON motion capture system

Compared with previously discussed motion capture systems in previously in Chapter 2 under subsection 2.2.5, MAC3D and VICON have been widely utilized in the research field especially related to human gait and posture. These systems comprise reliable data accuracy that makes use of an effective marker-based motion capture system. A recent review of the literature on Vicon [206] found that the system could be used to measure kinematic and kinetic parameters of the spine with good accuracy and reliability. Both systems share many things in common, such as the use of the three-dimensional coordinating system with a number of infrared cameras and the use of reflective spherical markers for the kinematic and kinetic analysis. The only differences are the manufacturer and software used during recording and analysis.

3.3.2 Participants

In total, twenty-one asymptomatic male subjects were recruited for this experiment. The subjects had the following characteristics: Average age of 24.67 (SD: 4.56) years and BMI of 22.99 kg/m² (SD: 3.78). All subjects were free from any health issues especially back pain symptoms for the preceding 6 months. In all cases,

subjects' consent was obtained. GPower analysis was done to guarantee that the study's minimal number of subjects was sufficient, as illustrated in Figure 3.2. One and Two-way ANOVA repeated measures tests were chosen for the statistical analysis and other relevant data were used as the inputs for the analysis. The minimal number of subjects for this study, based on the result of the analysis, is 18. Thus, the number of subjects who participated in this study was more than sufficient. Furthermore, Seerden, Dankaerts [208] and Ignasiak, R  ger [38] used 18 and 21 subjects, respectively, in similar studies relating to spinal kinematics done on healthy subjects.

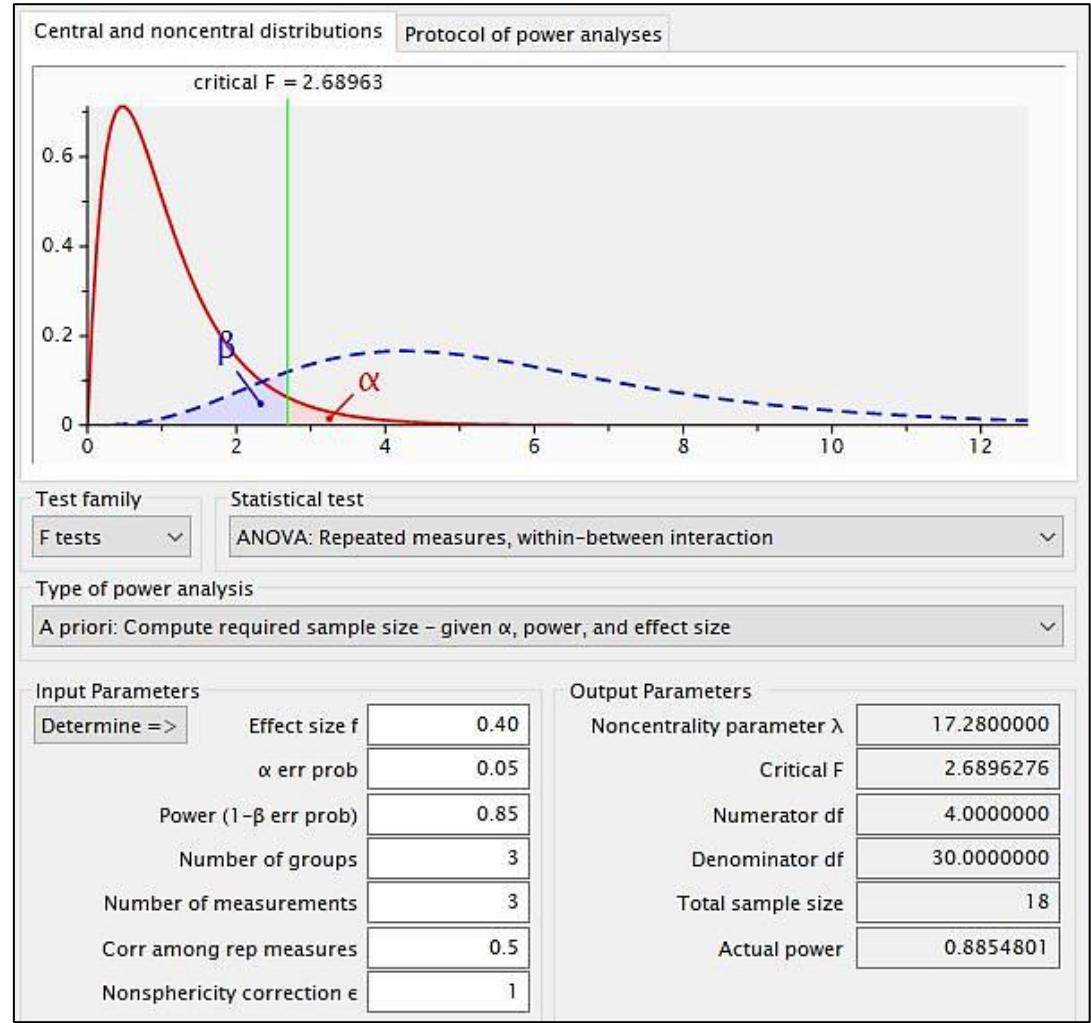


Figure 3. 2 Calculation of sample size using GPower

3.3.3 Instrumentation

The system used a 3-D motion capture analysis system with ten high-precision infrared cameras (HWK-200RT camera, Motion Analysis, USA). The system converts a reflective passive marker's real-time location to three-dimensional coordinates which are then used for kinematic analysis. The sampling rate was set at 200 [Hz]. A total number of 21 reflective passive markers with a diameter of 10 mm (Figure 3.3) were attached behind the spinous process of the spine (cervical: 3, thoracic: 12, lumbar: 5, and sacral 1) using tape as in Figure 3.4. The markers were also placed at both shoulders, waist, knee, and ankle of the subjects. A total of 29 markers were placed. The present study refined the method of cervical angle measurement put forward by [209] that implemented Cobb's method. All the markers were placed by a single certified physiotherapist to reduce the measurement error. Participants were asked to only wear a given cap and pants to avoid errors resulting from the movement of the skin [36, 210]. Four reference markers were placed on the test floor as reference points during data analysis. Before beginning the experiment, the system was calibrated using the Wand Calibration Capture Date function by requesting a participant to wave the hand for approximately 60 seconds until the camera image region becomes green and the numerical value fell within the acceptable range. Additional trials were conducted in case the numerical value fell outside the acceptable calibration range. For the STS task, an office chair, armless and backless, was adjusted vertically for each subject to obtain the same knee flexion angle (fixed at 90°).



Figure 3. 3 Dimension of reflective spherical marker used

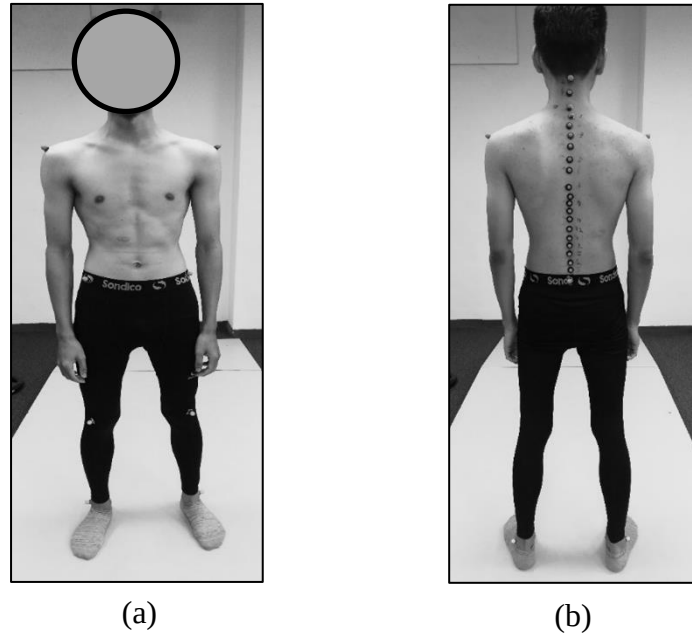


Figure 3. 4 Marker placement on the subject's body, (a) Back and (b) Front view

3.3.4 Experimental procedure

As mentioned in section 3.1, this research focuses on exploring the segmental kinematic of the spinal curves during critical daily activities. 13 different tasks for every subject have been analyzed by using this motion capture system. The details on the motions are as specified below:

(a) Sit to Stand

Each subject was asked to complete the task at self-selected speed with the feet self-positioned over the force platform (no fixed distance between the feet was imposed) for 3 acquisition trials. The time interval between each trial was fixed at 5s. The experimental setup is similar to the one proposed by Sibella, Galli [131]. The task begins from sitting (0% of motion completion) and ends at a quiet standing posture (100% of motion completion) as shown in Figure 3.5.

(b) Forward Bending from standing posture with three different approaches

Participants were instructed to perform forward bending movement which begins from a quiet standing posture at self-selected speed with the feet self-positioned over the force platform (no fixed distance between the feet was imposed) for 3 acquisition trials, with arm relaxed next to the thigh. This study adopted the method used by Tully, Fotoohabadi [108]. Participants were asked to practice the motions sequence two to three times to ensure all of them familiarize themselves with the activity. The movement was repeated if the examiner was not satisfied with the data such as the significant inconsistency in movement and speed of transition resulting from unwanted actions. Three different forward bending approaches were studied, namely Straight Hand (FBSH), Straight Back (FBSB), and Hanging Hand (FBHH) as in Figure 3.6. For FBSH and FBSB, both hands were placed on the knee to provide additional support for the back during the bending posture. The only difference was FBSH used maximum hand flexion while FBSB only uses minimal.

(c) Pick an object on the floor with three different approaches

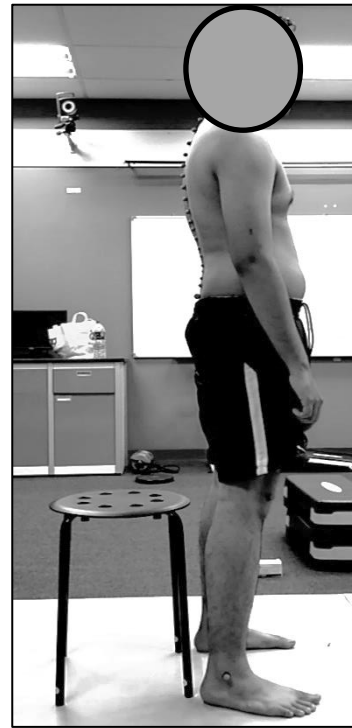
Participants were instructed to perform forward bending movement which begins from a quiet standing posture at self-selected speed with the feet self-positioned over the force platform (no fixed distance between the feet was imposed) for 3 acquisition trials, with arm relaxed next to the thigh. Participants were asked to practice the motions sequence two to three times to ensure all of them familiarize themselves with the activity. Three different pick object approaches were studied, namely Stoop (POST), Semi-squat (POSS), and Squat (POSQ) as in Figure 3.7. For POST, the motion at the knee was controlled and restricted while maximum flexion of the trunk was employed as the subject tried to reach and pick the object. During POSS, both trunk and knee are required to be flexed. The knee flexion was kept minimal and ensured it does not go beyond 90° . This approach utilizes lesser trunk flexion compared with POST. POSQ approach is the most recommend approach by physiotherapists in picking objects at a lower height as the trunk flexion is kept minimal while the knee flexion is maximized.

(d) Weightlifting with three different approaches

Participants were instructed to perform forward bending movement which begins from a quiet standing posture at self-selected speed with the feet self-positioned over the force platform (no fixed distance between the feet was imposed) for 3 acquisition trials, with arm relaxed next to the thigh. Participants were asked to practice the motions sequence two to three times to ensure all of them familiarize themselves with the activity. Similar to the previous task, this task requires the subject to perform weightlifting different loads of 5kg and 10kg with three different approaches, which are Stoop (ST), Semi-squat (SS), and Squat (SQ) as in Figure 3.8.

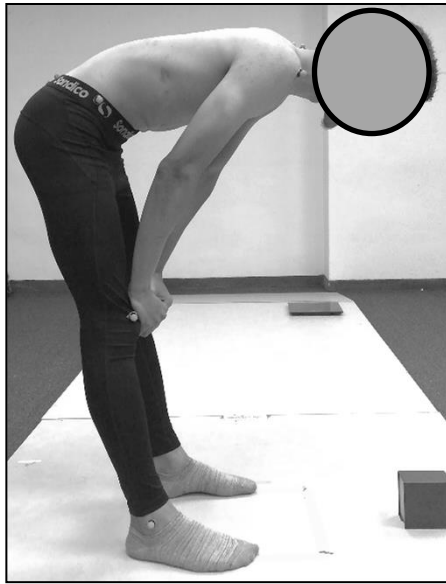


(a)

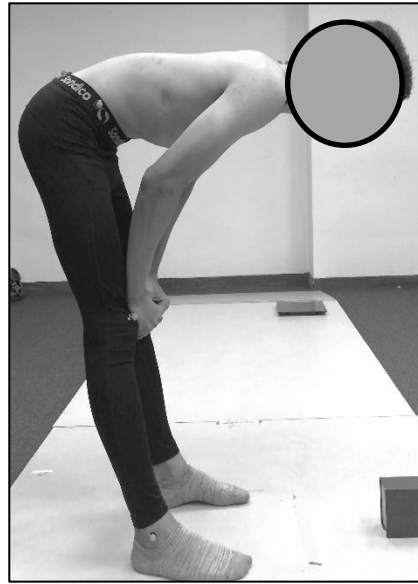


(b)

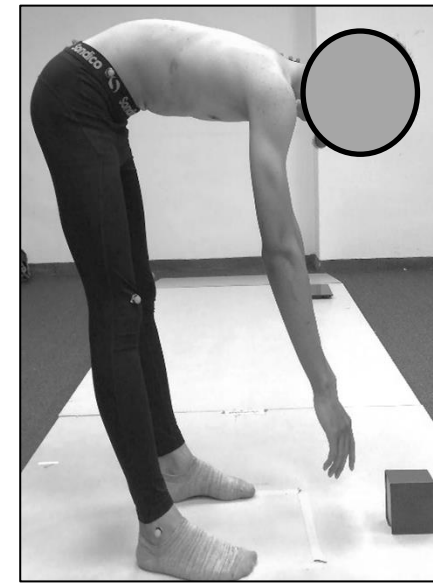
Figure 3. 5 Photo of the STS cycles. (a) Sit and (b) Stand



(a)

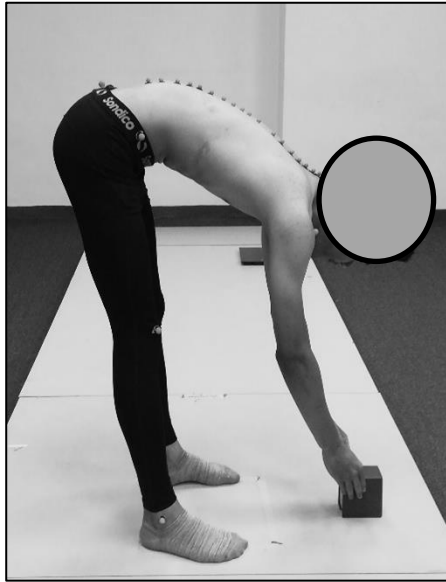


(b)



(c)

Figure 3. 6 Forward Bending with different approaches. (a) Straight Hand, (b) Straight Back, and (c) Hanging Hand



(a)

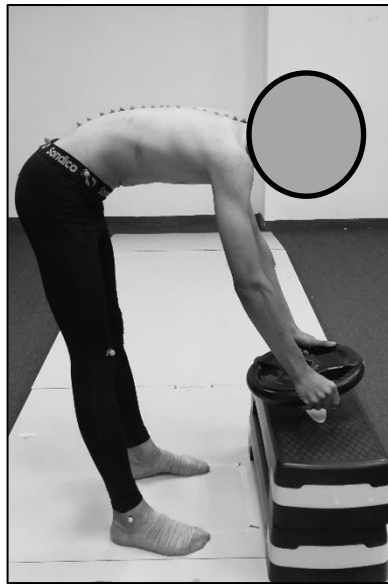


(b)



(c)

Figure 3. 7 Pick an object on the floor with different approaches. (a) Stoop, (b) Semi-squat, and (c) Squat



(a)



(b)



(c)

Figure 3. 8 Weightlift a load on the floor with different approaches. (a) Stoop, (b) Semi-squat, and (c) Squat

Table 3. 1 Comparison of motion capture systems used for data collection in this study

Instrument used	Site	Complexity in handling	Software used	Frequency	Detection object	Advantages	Disadvantages
Stick-markers based system	Within UTM	Medium	MATLAB and Microsoft Excel	240 Hz	Stick-markers	• Easy to handle • Low-cost measuring system.	• A limited number of similar studies • Require manual angle measurement. Requires assistance of physiotherapists to palpate spinal landmarks.
KINECT Xbox Skeleton Tracking		Low	KINECT for Windows	25–35 Hz	Human movement (Marker-less)	• Easy to handle • No markers needed	• Only up to 2 spinal points can be tracked. • Has compatibility issues due to the format's not interchangeable.
3D Investigator		Medium	NDI First Principles	4600 Hz	Wire-based sensors	• - Quick coordinate measurement system.	• Difficult to handle due to stiff cables and robustness of the sensor's orientation. • Each sensor must be connected to two wires at both ends. • Cannot be placed on many spinal segments
Vicon		Medium	Nexus	200 Hz	Reflective spherical markers	• Quick and reliable coordinate measurement system. • Easy to handle	• Only up to 2 spinal points can be tracked.
MAC3D	Shibaura Institute of Technology, Japan	Medium	Cortex	200 Hz		• Easy to handle. • Accurate markers positioning	• A limited number of similar studies for spinal multisegment.

3.4 Analysis

3.4.1 Data Analysis

Data processing was performed by using specified software (Cortex version 6) then converted to an Excel file. Measured data were collected in terms of three-dimensional coordinates. The raw data were filtered by a second-order Butterworth filter with a 6Hz cut-off to eliminate noise especially from the power line and movement. Customized MATLAB codes (Matlab 2016b, Chicago IL, USA) were used for data analysis. From the processed data, the placement of markers on the subjects' bodies was compared with literature that used X-ray and radiography approaches [211-214] by using the image processing method at a standing posture to make sure the vertebra locations were correctly assumed. Van Blommestein, MaCrae [60] stated that standing radiographs are the gold standard method for measuring cervical, thoracic and lumbar angles. Thus, the comparison study makes use of MRI and Radiography studies of standing posture.

3.4.2 Global and segmental calculations

Similar approaches were used to calculate global and segmental spinal angular displacement. The limit between the thoracic and lumbar region was set on the 10th thoracic vertebra based on variation in facet joint orientation [215] and radiographs of standing posture [31]. The thoracic region is between T1 to T10 while the lumbar region is between T10 to L5. The angular displacement of the global thoracic curve was measured between the straight lines of T1-3 and T8-10. For the global lumbar curve, the relative angle is between the straight lines T10-12 and L3-5 (Figure 3.9). For the segmental analysis, thoracic and lumbar regions were divided into 4 (T1-4, T4-6, T6-8, and T8-10) and 3 (T10-L1, L1-3, and L3-5) segments, respectively. The angular displacement of the segments was derived using a 3-point coordinate system in MATLAB. For example, for segment T10-12, the coordinates of markers on the 10th, 11th, and 12th thoracic were used. The mathematical analysis on measuring the angular displacement was loosely based on the one proposed by Tully, Fotoohabadi

[108], and Claus, Hides [101]. From the results, the mean and standard deviation of angular displacement was calculated for 0 to 100% of motion completion for both cycles of all tasks. Each participant displayed a difference in time (second) of completion for all tasks. Thus, to normalize the range of initial and end of motion completion, a percentage scale was used in this research.

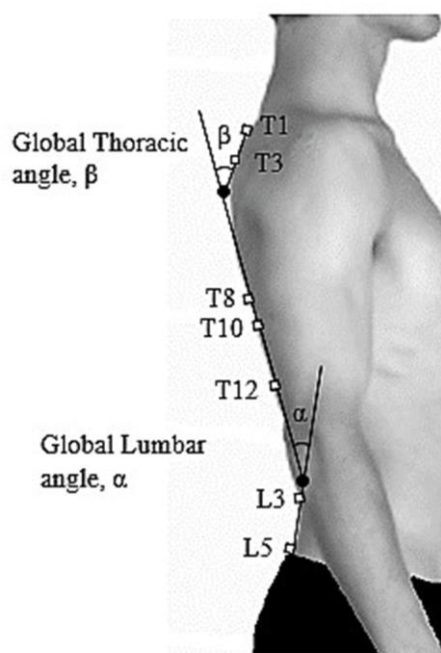


Figure 3. 9 Measurement of global thoracic and lumbar angles.

3.4.3 Mean and Segmental differences

For the analysis, all results were normalized to standardize the discrepancy of angular displacement values between subjects. The initial value of the angular displacement for the Begin cycle was set to 0. For the Return cycle, the end values were set to 0 to demonstrate the relationship between both cycles in each task studied. The span between the beginning and end of these cycles was then divided into ten smaller phases to improve the visibility of kinematic changes as the task progressed. The analysis for the obtained angular displacements was then divided into two different calculations, which were the Main (Equation 3.1) and Segmental Differences (Equation 3.2). The “main difference” calculation was used to measure the range and changes of the segmental kinematics of flexion-extension of the Begin and Return

cycles. Equation 3.2 is the modification of the earlier equation to highlight the changes of kinematics specifically in every phase. This equations and normalization approach used in this study is adapted from [38, 101, 108] and [216].

$$\text{Mean difference} = \text{Current phase} - \text{Initial phase} \quad (3.1)$$

$$\text{Segmental difference} = \text{Current phase} - \text{Previous phase} \quad (3.2)$$

3.4.4 Hip and Knee Angle

As mentioned in subsection MAC3D and Vicon, the markers were also placed at both shoulders, waist, knee, and ankle. These markers were used to calculate the hip and knee angle as illustrated in Figure 3.10 below. For the hip angle, the angle was calculated between markers placed on the shoulder, waist, and knee. The knee angle was calculated between markers placed on the waist, knee, and ankle. The waist and knee markers were taken as the center of rotation for hip and knee angle calculations, respectively.

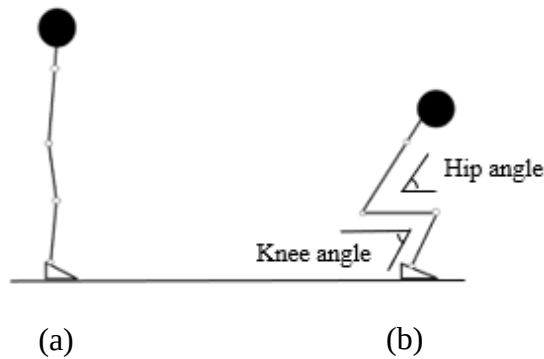


Figure 3. 10 Hip and Knee angle measurement used. (a) Stand and (b) Semi Squat postures

3.4.5 Statistical analysis

The mean and standard deviation of all angular displacement was calculated for all tasks studied. Statistical differences for the mean of angular displacement in different phases between global and segmental measurements for both thoracic and lumbar were calculated using one-way ANOVA. To study the interaction between approaches of a similar task group, a two-way ANOVA analysis was conducted. For statistical analysis, results were analyzed using MATLAB version 2018a with an alpha level of 0.05. The reliability of the angular displacement and kinematics of the cervical, thoracic, and lumbar was assessed using intraclass correlation coefficients (ICC). The method for proper selection of ICC analysis given in [202] is comprehensive and has been used for these studies. ICC estimates and their 95% confidence intervals were calculated using SPSS statistical package version 22 (SPSS Inc, Chicago, IL) based on a mean-rating ($k = 3$), absolute-agreement, 2-way mixed-effects model. Shrout and Fleiss [205] and Koo and Li [202] call into question some past agreement of the ICC value above 0.75 shows excellent agreement between measured variables. Table 2 shows the general guideline proposed by Koo and Li [202] for better interpretation of ICC's values:

Table 3. 2 General interpretation of ICC value

ICC value	Reliability's interpretation
< 0.5	Poor
$0.5 < \text{ICC} < 0.75$	Moderate
$0.75 < \text{ICC} < 0.9$	Good
> 0.90	Excellent

3.5 Palpation techniques implemented

As previously discussed in Chapter 2 under subsection 2.2.6, there are several techniques used to palpate different spinal curves. The bony landmarks are the basic and common guide used for the palpation of spinous processes within the cervical, thoracic, and lumbar regions. For cervical, the conventional and flexion-extension/motion are the most common methods used, which are also being implemented in this study. An additional palpation method was used which used olecranon and styloid processes of Ulna as references for the lumbar region. The details of the palpation approach used in this study were discussed in detail in the next subsection.

3.5.1 Palpation by bony landmarks

To correctly identify the spinal vertebra that is made up of many segments, bony landmarks or surface anatomy are used as references. However, it is important to take note that only specific vertebra can be correctly identified using this approach while others need to be identified by running hands over. Vertebrae that can be identified using bony landmarks are C7, T3, T7, L4, and S2 as illustrated in Figure 3.11. The third and seventh thoracic spinous processes are identified using the subject's scapula. T3 is assumed in line with the root of the spine of the scapula (RSS) while T7 is located within the inferior angle of the scapula (IAS). For the lumbar, the palpation by bony landmarks can only estimate the position of L4, which is in line with the highest part of the iliac crest. S2 is found within the posterior superior iliac spine (PSIS) and located second from the top using palpation.

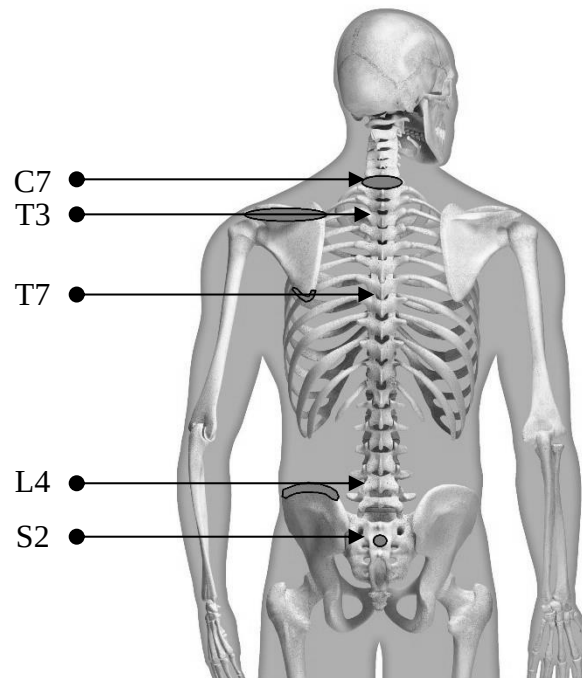


Figure 3. 11 Vertebra that can be identified using bony landmarks

3.5.2 Lowest cervical vertebra as reference

This study makes use of a combination of both static and motion palpation techniques for cervical. This approach is geared toward the findings highlighted by Shin, Yoon [92], and Merz, Wolf [90] which found the combination of approaches can increase the level of accuracy.

Three fingers are used as the initial assumption for the cervical spinous process as in Figure 3.12 below. Look for the most prominent cervical spinous process and assume it as C7. Investigator's or physiotherapist's index and middle finger are placed on top of the spinous process of C6 and C7, respectively, with the seated subject's cervical spine in flexion. The subject is then asked to perform a cervical extension movement while the fingers are maintained. If the index finger moved anteriorly while the middle finger remained stationary, the lower cervical spinous process would be labeled as C7. However, if both fingers stayed stationary, the top cervical palpated would be defined as C7, and the procedure will be repeated by shifting 1 level cephalad at a time to validate C7's location.

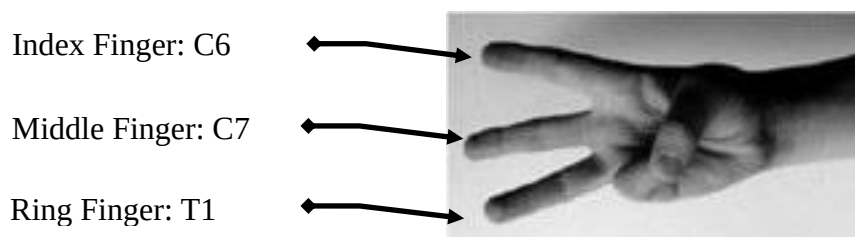


Figure 3. 12 Placement of fingers on the cervical spinous processes

3.5.3 Olecranon and styloid processes of Ulna as landmarks

The length of the Ulna is used to estimate the location of the upper and lower lumbar vertebra. A straight line is marked at the wrist as the subject performs wrist extension as shown in Figure 3.13. Also, mark with the 'X' symbol behind the elbow (olecranon process) and at the middle between the elbow and the straight line at the wrist. Locate the subject's coccyx by finding the lowest part of his/her spine. The straight line marked on the subject's wrist must be in line with the location of the subject's coccyx as shown in Figure 3.14. L1 location is estimated in line with the 'X' located on the subject's elbow while the L5 is in the middle point between the elbow and the line marked on the wrist. Then, the subject is required to be prone with a small pillow below the chest. The location of the spinous processes of the vertebra is then palpated and marked using a marker.



Figure 3.13 Wrist extension and location of the styloid process (circled)

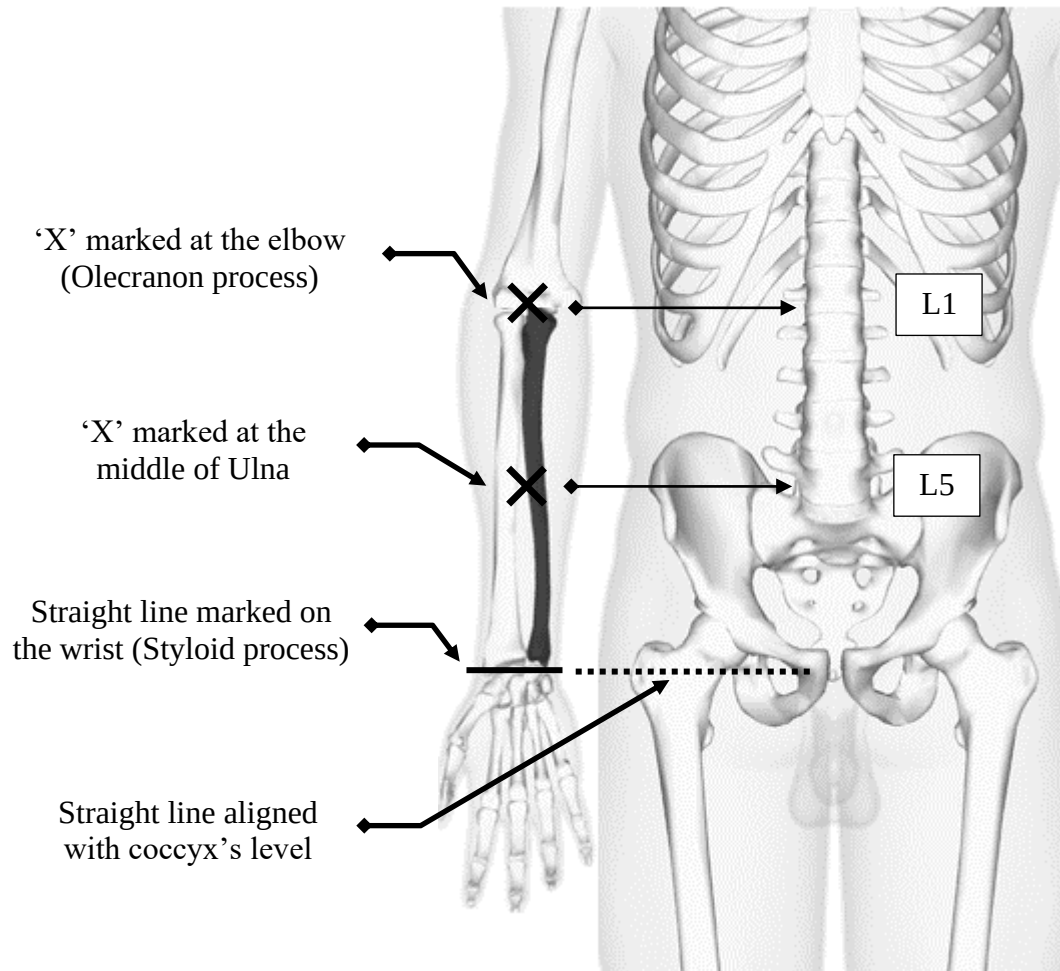


Figure 3.14 Estimating lumbar endpoints using Ulna bone (colored in dark grey)

CHAPTER 4

ANALYSIS OF SPINE

4.1 Introduction

To highlight and understand the vertebral contribution within the cervical, thoracic, and lumbar spine and their correlation in different daily tasks, a number of studies have been conducted using different approaches. The details for the participants, instrumentation, procedures and analysis used are discussed in detail in Chapter 3. The studies implement non-invasive approaches such as stick marker-based methods and reflective markers in understanding the global and multi-segments of the human spine within the cervical and lumbar curves. The results have also been critically discussed and compared with findings from the literature.

4.2 Approaches of spinal kinematics using MAC3D and VICON

4.2.1 Lumbar as a single and multi-segment

This section defines the lumbar region as a single and multi-segment during critical daily tasks of STS and Stand-to-flexion (STF). The assumption of the lumbar as a single segment is widely used by literature and already discussed in Chapter 2 under the subtopic Spine Biomechanics. As previously mentioned in Chapter 3 under subsection 3.4.2, this study defines T10 to L5 as the endpoints of the lumbar region when defined as a single segment. This segment is also defined as a global lumbar for this study. For multi-segment analysis, the lumbar was divided into three segments. The boundaries between these segments were defined as T10-L1, L1-3, and L3-5. The kinematic differences of the lumbar region between single and multi-segment during both tasks were also discussed in detail in this section.

4.2.2 STS: Lumbar as a single segment

The change of angular displacement for the global lumbar curve during STS is significantly different when studied as different segments compared to an individual segment (Table 4.3). Most subjects implemented the kyphotic lumbar curve at the initial phase ($8.38 \pm 7.93^\circ$) which was due to the absence of lower back support. The curve then gradually decreases to a lordotic curve until reaching 100% of motion completion ($-11.71 \pm 4.80^\circ$) as elucidated in Figure 4.1(a). Lord et al. noted that lumbar lordosis during sitting was nearly 50% lower on average as compared to standing lumbar lordosis[100]. Our results do not appear to validate the finding of Lord et al. In fact, the results showed approximately 70% on average standing lordosis compared to Lords' finding of 50%. The use of asymptomatic subjects in the present study could well be responsible for this result, as Lord et al. focused on reporting subjects suffering from back pain. An "ideal" form of lumbar transition from kyphotic to lordotic curve, which can be observed, occurred at 50% of motion completion. The lumbar spine begins to extend during the buttock's lift-off, which resulted in a higher lumbar spine curve's value at the end (100%) compared to the initial phase (0%).

4.2.3 STS: Lumbar multi-segment

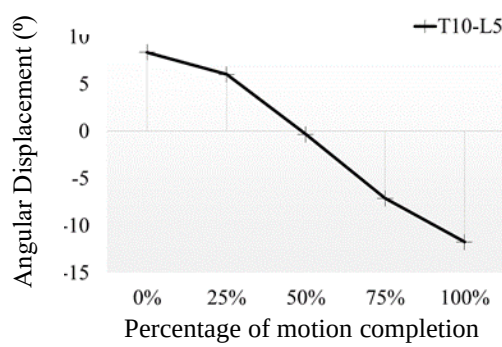
The statistical study revealed that there are no significant differences between the global and multi-segment using one-way ANOVA analysis as in Table 4.1 (P-value = 0.425). However, the lumbar multi-segments showed a statistical significance difference when analyzed without the global lumbar (P-value = 0.01). Carr highlighted the segmental contribution of the spine during standing up. However, the biomechanical analysis indicated that movement at the spinal joints is minimal [112]. The present study focused on highlighting the segmental contribution of the lumbar spine during STS since there is a lack of literature exploring it. From Figure 4.1(b), the upper segment (T10-L1) played a vital inclination point from the kyphotic to the lordotic curve for the lumbar region as it is located within the thoracolumbar joint. This segment only showed a peak increment towards the lordosis curve at 50% of motion completion and maintained passive kinematic with minimal changes (Δ less than 2°) in angle within the early and end of motion completion. Interestingly, this

segment showed the slowest motion transition (kyphotic to lordotic) compared with the other two segments and concluded to be passive before and after 50% of the motion completion phase due to its poor contribution during the transition. Presumably, the passive kinematic of this segment was due to the extension of the thoracic spine along with the flexion of the lumbar spine during STS [108]. The transition to lordotic kinematic for this segment only occurred after 75% of motion completion, thus validating point T10 as the top inclination point of the lordosis curve.

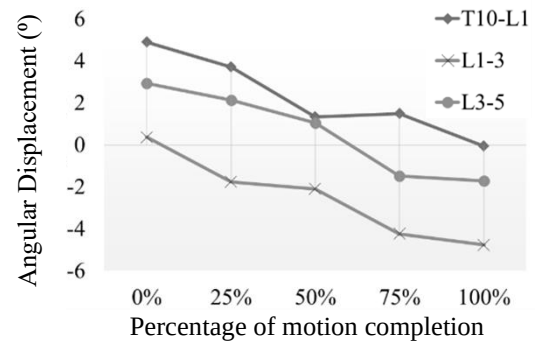
During early transition (0-25%), the middle lumbar segment (L1-3) tends to show early contribution during the transition with the highest angular displacement increment (Δ more than 2°) as in Figure 4.1(a). Compared to the other two segments, this segment showed the quickest response to change from kyphotic to lordotic curve, which occurred before reaching 25% of motion completion. This segment contributed more than 50% of the lumbar lordosis angle at the end of the motion. The result concluded that this part is active and has the highest mobility during STS. Latter response can be observed within the lowest segment (L3-5) compared to the middle segment, up until reaching 50% of motion completion with minimal changes (Δ less than 2°). The results showed that the passive/delayed kinematic of the L3-5 segment within this period (0-50%) and the peak of motion can be observed after 50% of motion completion as shown in Figure 4.2. The delay of angular displacement increment within this segment appears to indicate the translation from horizontal to vertical momentum [112]. The translation is necessary to move the upper body mass, which includes trunk, torso, and head mass, forward to the feet. The translation is essential as it provides initial stabilization which results in a delay of response within the lower lumbar segment. The peak increment of angle only occurred at 75% of motion completion and was maintained until the end of the motion. All three segments displayed different time responses based on anatomical and functional reasons during the transition from kyphotic to lordotic curve with the middle segment showing the quickest transition.

Table 4. 1 Change in lumbar's angular displacement during STS

Lumbar Segment	Motion Percentage					P-value	
	0%	25%	50%	75%	100%	W Global	W/O Global
T10-L5	8.38±7.93	6.02±7.29	-0.33±7.84	-7.11±6.90	-11.71±4.80	0.425	0.010
T10-L1	4.90±6.04	3.72±5.80	1.34±4.84	1.50±5.91	-0.05±5.17		
L1-3	0.38±5.12	-1.75±7.03	-2.09±5.87	-4.23±5.82	-4.76±5.08		
L3-5	2.94±3.92	2.13±4.41	1.05±3.98	-1.49±4.08	-1.71±3.44		



(a)



(b)

Figure 4. 1 Lumbar transition during STS. (a) Single and (b) Multi-segment

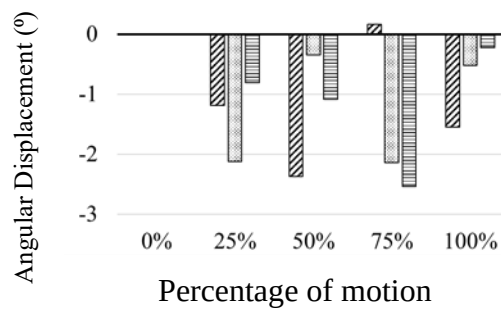


Figure 4. 2 Change of segmental angular displacement during STS

4.2.4 STF: Lumbar as a single segment

The transition from the lordotic to the kyphotic curve of the lumbar region was completed before achieving 25% of motion completion as shown in Figure 4.3(a). Linear increment towards the kyphotic curve was observed at the beginning of motion until reaching 50% of motion completion. The increment in motion speed tends to slow down upon reaching 50% of motion completion. Upon reaching 75% of motion completion, no change in angle was observed and it was maintained until reaching 100% of motion completion. The initial (0%) to intermediate (50%) motion transition showed a significant change in the global lordosis angle. It was difficult to highlight the actual vertebra contributing to the change if the lumbar was only studied as a single segment. Also, the results discovered delayed or passive kinematic of global lumbar acting between 50% to 100% of motion transition.

4.2.5 STF: Lumbar multi-segment

The statistical analysis revealed no significant difference was found between the lumbar global and multi-segment. Surprisingly, the top lumbar segment was found to be in flexion kinematic while other segments showed extension curves as shown in Table 4.2. The upper lumbar segment (T10-L1) showed passive kinematic with minimal changes (Δ less than 2°) in angular displacement compared to the middle and lower lumbar segments. This segment was concluded to be passive and has the lowest contribution during flexion maneuvers. The delayed and passive kinematic of this segment was due to the anatomical location as this segment serves as the connection that linked the thoracic kyphosis and lumbar lordosis or known as thoracolumbar. The thoracic curve was not affected as much as the lumbar curve during the forward bending, thus resulting in the passive kinematic of the peak lumbar segment.

Middle (L1-3) and lower (L3-5) lumbar segments displayed approximately similar kinematic but a quicker change in angular displacement was observed within the lower lumbar segment as in Figure 4.3(b). Interestingly, the peak of change in

angular displacement of the middle segment was observed specifically at 25% to 75% of motion completion. As a result of its moderate contribution to the motion transition, this segment was considered to be an active segment. The lower segment (L3-5) had more mobility than the middle segment (L3-5), which had a slower temporal response. At the beginning of motion, the patterns of the lower lumbar segment and global lumbar lordosis were nearly identical (0-25%). A gradual increment of angular displacement was observed throughout task completion within this segment as in Figure 4.4. At 50% of motion completion, the peak angular displacement increment was also observed, demonstrating the importance of the lower segment in the overall change in lumbar lordosis and concluding that it is an active segment. The high mobility kinematic displayed by the lower lumbar region is barely distinguishable from Alqhtani, Jones [128] who found strong correlations between the lower lumbar (L3-S1) spine's range of motion (ROM) during flexion.

Table 4. 2 Change in lumbar's angular displacement during STF

Lumbar Segment	Motion Percentage					P-value	
	0%	25%	50%	75%	100%	W Global	W/O Global
T10-L5	-6.46±7.82	6.97±8.94	21.16±8.87	25.69±7.68	26.16±7.92	0.066	0.792
T10-L1	2.07±3.43	3.72±2.32	4.39±2.78	2.86±5.13	3.24±4.24		
L1-3	-0.44±2.86	0.87±3.86	3.97±3.58	5.35±2.31	4.82±3.23		
L3-5	-1.11±4.68	2.17±3.62	6.26±2.15	7.92±2.04	4.82±2.73		

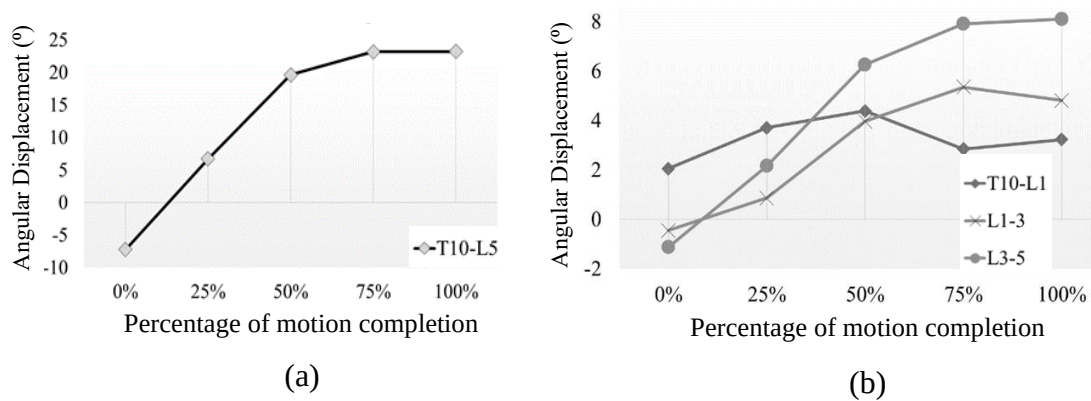


Figure 4.3 Segmental lumbar transition during STF. (a) Single and (b) Multi-segment
*Positive and negative value indicate kyphotic and lordotic lumbar curve, respectively.

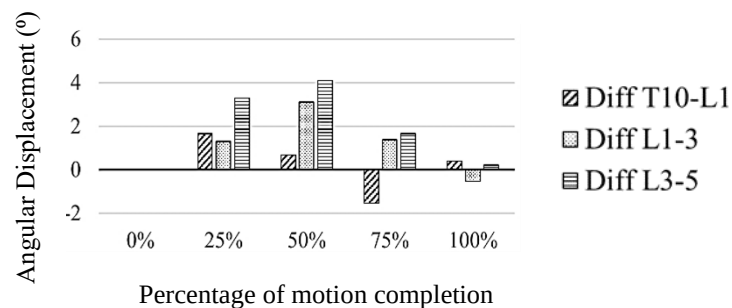


Figure 4.4 Change of segmental angular displacement during STF

4.3 Contribution of cervical and thoracic multi-segment during STS & STF

This section defines the cervical as one segment and thoracic as five small segments during critical daily tasks of STS and STF. The boundaries used for the Cervical region were C2 to C7. For thoracic, these segments were defined as T1-4, T4-6, T6-8, and T10-12. The kinematic differences of these regions during both tasks were also discussed in detail in this section.

4.3.1 Kinematics and reliability of the spinal region

The repeatability and reliability of the cervical and multi-segment thoracic spine were assessed using the interclass correlation coefficient (ICC) for both tasks specified during the study (Table 4.3). The single most striking observation to emerge from the result was the excellent reliability with all ICC values above 0.90 (ranged between 0.932-0.976). Results for the statistical study of cervical and thoracic segments during STS and STF showed different findings. There were statistically significant differences for the STS task at all levels and the initial and end of STF task with an exception at other levels as shown in Table 4.4. The averaged angular displacement and standard deviation at each phase are presented in Table 4.5 (STS) and 4.6 (STF). The range of cervical flexion and extension obtained in this study during STS and STF were within the acceptable range of cervical flexion compared with previous studies that used [125] Optical motion capture system (F:58.2°±12.6, E:55.5°±10.0), [126] ultrasound-based coordinate measuring system, CMS (F:52.2°±8.9, E:76.5°±16.6) and Goniometer (F:52.9°±8.1, E:79.1°±17.0).

Table 4. 3 Intraclass correlation coefficient (95% CI) of the cervical and multi-segment thoracic region during STS and STF tasks.

	Segment	ICC	95% CI
STS	C2-T1	0.944	(0.820,0.993)
	T1-4	0.961	(0.864,0.995)
	T4-6	0.956	(0.845,0.995)
	T6-8	0.970	(0.905,0.996)
	T8-10	0.975	(0.921,0.997)
	T10-12	0.969	(0.902,0.996)
STF	C2-T1	0.932	(0.743,0.992)
	T1-4	0.973	(0.896,0.997)
	T4-6	0.976	(0.908,0.997)
	T6-8	0.965	(0.864,0.996)
	T8-10	0.968	(0.873,0.996)
	T10-12	0.964	(0.827,0.996)

Table 4. 4 P-value for the segmental angle of the cervical and thoracic spine during STS and STF

Percentage	STS	STF
0%	<i>P-value <0.05</i>	<i>P-value <0.05</i>
25%	<i>P-value <0.05</i>	0.14
50%	<i>P-value <0.05</i>	0.51
75%	<i>P-value <0.05</i>	0.14
100%	<i>P-value <0.05</i>	<i>P-value <0.05</i>

4.3.2 STS

The results in Table 4.7 revealed the difference of multi-segmental kinematics (change in angular displacement) within cervical and thoracic regions. This finding substantiates previous findings in the literature [217] with coordinate variation patterns that suggested differences in regional movement timing within the thoracic spine. For the global thoracic angle, the sitting posture ($130.09^{\circ} \pm 12.02$) compress the intervertebral disk, thus resulting in lower angular displacement compared to upright standing ($150.04^{\circ} \pm 6.50$). The results from Figure 4.5(a) revealed the transition from flexion to the extension of angular displacement within all segments (cervical and thoracic T1-12). Statistical results also revealed that there are significant differences between cervical and thoracic multi-segments during the STS task.

The transition of flexion-extension occurred approximately in the middle of STS with the cervical region displaying a quicker transition. There was a trend towards an increase of extension following standing. The present study is consistent with a previous study [218] which found the dominance of the extension compared with flexion during the STS task. Although Fotoohabadi, Tully [219] found the extension of the thoracic spine begins at 30.12% of STS transition, the present study found the transition occurred at approximately 50%. This would appear to indicate that age difference tends to manipulate the flexion-extension of the spinal region as Fotoohabadi et al. [219] used older subjects in their study. At this point (middle STS phase), spinal compression starts to take place as it needs to move upper body weight

vertically. A previous study by [220] supported these findings which found the maximum compressive loads at middle-lower thoracic levels occurred at half the task time of the STS transitions. A minimal increment of angular displacement was noted within the cervical spine, which may be due to the cervical lordotic anatomical curve which is the same as the lumbar spine. Wang, Lindstroem [221] also affirm that cervical joints repeat their motion accurately between days for flexion and extension motions.

The top thoracic segment (T1-4) displayed a minimal change of angular displacement compared with other thoracic segments. The peak of increment was observed at 75% of motion completion at all segments including the cervical region as in Figure 4.5 (b). The least change in angle increment of flexion-extension was observed at the middle (50%) of STS transition, in which the compression within the anterior region of the cervical and thoracic spine takes place, which also highlighted the passive condition. Upon reaching the standing posture, the least increment occurred within the lowest thoracic segment, T10-12. Other thoracic segments displayed the approximately equivalent percentage of contribution toward upright standing. The lowest thoracic segment (T10-12) showed the least increment within 75-100%, which is due to the anatomical structure at the thoracolumbar joint and the nearest to the lumbar region. These (lower thoracic and upper lumbar) regions are responsible to support upper body weight (torso, head, hands, etc.), resulting in lower angular displacement increment and variation in movement during upright stand.

Table 4. 5 Normalized angular displacement (SD) of the cervical and thoracic multi-segment during the STS task.

Segment	Percentage of motion completion			
	25%	50%	75%	100%
C2-T1	-15.32±7.39	-14.29±17.25	10.43±9.94	27.66±12.15
T1-4	-27.93±12.33	-34.63±17.21	3.49±16.61	31.07±15.45
T4-6	-30.55±14.42	-39.99±22.84	4.56±19.07	32.54±14.48
T6-8	-31.91±15.87	-43.35±23.18	3.07±15.89	26.26±15.00
T8-10	-33.65±14.97	-43.53±23.13	2.97±16.64	20.80±12.27
T10-12	-34.35±15.48	-45.09±23.72	-1.02±16.12	6.60±11.64

*negative value indicates flexion direction

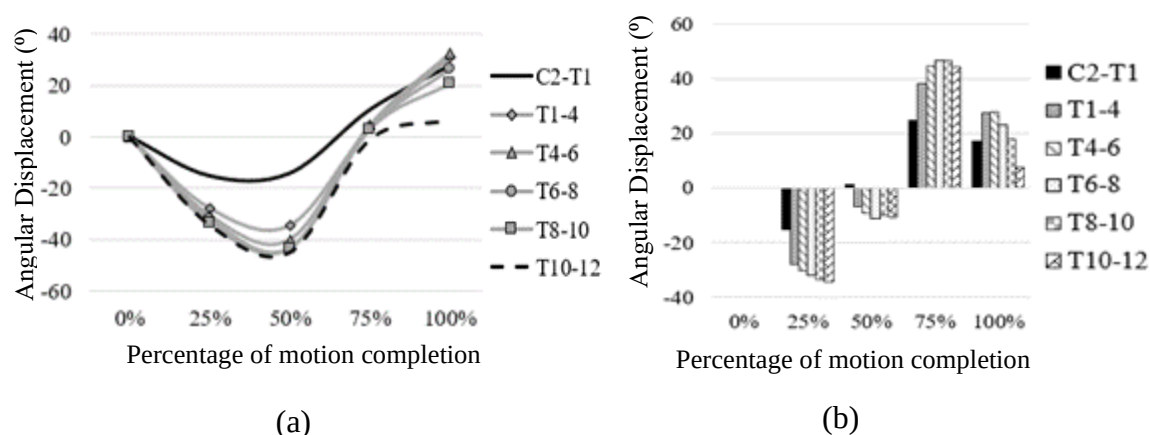


Figure 4. 5 Kinematics of the cervical and thoracic spine during STS (a) Normalized mean angular displacement and (b) Segmental Increment

4.3.3 STF

The Cervical and the top thoracic region had the least contribution during STF compared with the other thoracic segments specifically after the middle transition (Table 4.8). Compared with both segments, other lumbar segments displayed a greater difference in angular displacement, especially segments T8-10 and T10-12. This result has further strengthened our findings that the thoracic multi segments did not share the same kinematic of flexion-extension during daily STS and STF tasks. In contrast to

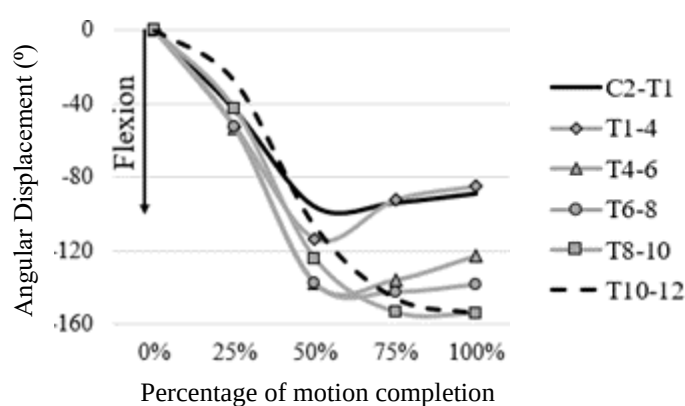
earlier findings [88], we did not find the upper thoracic segment (T1-4) to have a higher contribution during STF compared with the other lower segments (T4-12). Curiously, the results also revealed the conversion of flexion-extension which occurred only within cervical and T1-8 segments approximately in the middle of the STF motion transition. The study by Bible, Biswas [218] found the forward bending at the waist has the greatest cervical sagittal motion recorded, compared with other daily movements (i.e. STS, walking, reading, etc.), thus indicating the importance of implementation of good posture while bending such as squatting to avoid progressive neck pain.

Statistical results found significant differences only at the initial and end of the STF movement. During STF, most of the segments reach the peak of flexion at a 50% phase with lower thoracic segments (T8-12) showing latter kinematic achieved at a 75% phase (Figure 4.6 (a)). Inexplicably, this finding highlights the occurrence of passive kinematic within the cervical and upper thoracic segments (T1-8) that begin at the middle while the lower thoracic segments (T8-12) begin at 75% of STF transition. Interestingly, the spinal segments in this study showed identical curve patterns with adjacent segments. The curve characteristic of the top thoracic segment (T1-4) displayed approximately similar kinematic with the cervical region. The middle thoracic segments (T4-6 and T6-8) showed an approximately identical curve with greater extension that occurred within the T6-8 segment. T8-10 and T10-12 segments shared identical curve patterns with T8-10 showing a higher change in angular displacement. Cervical and upper thoracic segments (T1-4) showed the least segmental flexion-extension increment throughout the STF task especially after reaching the middle STF transition as presented in Figure 4.6 (b). A study by [222] found that the static flexion period within the cervical spine should be shortened as it may result in an interruption in the flexion-relaxation phenomenon. It is also important to lengthen the activation of the cervical erector spinae muscles by increasing the frequency of activity and movement including the neck. Peak increment of flexion was observed within early motion transition (25-50%).

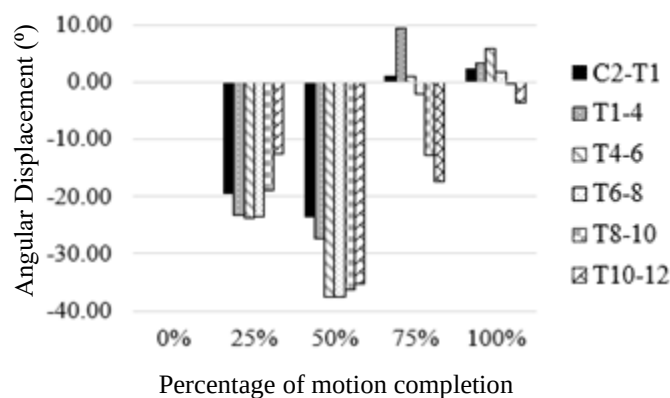
Table 4. 6 Normalized angular displacement (SD) of the cervical and thoracic multi-segment during the STF task.

Segment	Percentage of motion completion			
	25%	50%	75%	100%
C2-T1	-19.36±13.26	-42.82±11.10	-41.79±1.66	-39.58±31.16
T1-4	-23.33±13.82	-50.55±11.29	-41.12±6.60	-37.83±30.82
T4-6	-23.67±21.38	-61.13±17.72	-60.26±3.61	-54.48±45.56
T6-8	-23.52±22.36	-61.16±19.25	-63.28±1.14	-61.51±48.94
T8-10	-19.02±25.48	-55.30±23.31	-68.15±7.53	-68.52±54.17
T10-12	-12.50±26.76	-47.72±25.66	-65.02±11.19	-68.67±54.07

*negative value indicates flexion direction



(a)



(b)

Figure 4. 6 Kinematics of the cervical and thoracic spine during STF
(a) Normalized mean angular displacement and (b) Segmental Increment

The main purpose of this study is to highlight the differences in kinematic contribution during critical activities that are routinely used to complete daily tasks. Activities that are included in this study are forward bending, pick an object on the floor, and weightlift with 5kg and 10kg of load. According to [28], flexion is the most often utilized action daily, with up to 4400 flexions possible. As a result, the flexion motion is included and it is approached differently, which is with Straight Back (FBSB), Straight Hand (FBSH), and Hanging Hand (FBHH). For FBSB, the subject has been instructed to bend the trunk forward with flexion in both the thoracic and lumbar spine. By doing so, the angle between the trunk and the knee will be approximately 90°, when viewed from a coronal or side plane. On the other hand, the thoracic and lumbar spine are in a more relaxed posture with no postural control being applied, during FBSH compared with FBSB. For these tasks, both hands were placed on the knee to provide additional support for the back during the bending posture. The only difference was FBSH used maximum hand flexion while FBSB only uses minimal. In contrast to FBSH and FBSB, the hand is instructed to hang freely for the FBHH task.

For pick an object on the floor and weightlifting tasks, the motions were approached in three different ways, which are Stoop, Semi-squat, and Squat. During the pick object task, subjects were instructed to pick an object with a small mass from the floor. This study only focused on the effects of different postural approaches to the spine, thus the effect of the object's mass was neglected, compared with the weightlift tasks. For Stoop, the motion at the knee was controlled with restricted flexion while maximum flexion of the trunk was employed as the subject tried to reach and pick the object or load. During Semi-squat, both trunk and knee were required to be flexed. The knee flexion was kept minimal and ensured it does not go beyond 90°. This approach utilizes lesser trunk flexion compared with Stoop. The squat approach is the most recommend approach by physiotherapists in picking objects or carry loads at a lower height as the trunk flexion is kept minimal while the knee flexion is maximized. Thus, the effort of back muscles is greatly reduced, preventing lower back pain in the long run. To avoid confusion between approaches used for both pick an object and weightlifting tasks, these tasks are then renamed. For pick an object task, it is renamed to POST, POSS, and POSQ for Stoop, Semi-Squat, and Squat approaches, respectively. The tasks for 5kg and 10kg weightlifting are renamed according to the

load category. WL5ST, WL5SS, and WL5SQ stand for Stoop, Semi-Squat, and Squat methods, respectively, for the 5kg weightlift task. The 10kg weightlift task is renamed as WL10ST, WL10SS, and WL10SQ, which stand for Stoop, Semi-Squat, and Squat, respectively.

4.4.1 Result

The mean of angular displacement (with SD) for Mean and Segmental difference across all the critical activities studied were calculated and analyzed. More details on these results are in Appendix G (Forward Bending), H (Pick an object on the floor), I (Weightlift 5kg of load), and J (Weightlift 10kg of load). These results were obtained using the equation as previously mentioned in Chapter 3 under subtopic “Formula used to calculate the Main and Segmental differences”. Line graphs of normalized "Mean" differences and stacked bar charts for "Segmental" differences were created using the data from the tables to portray the findings correctly.

4.4.1.1 The intraclass correlation coefficient, ICC

An interclass correlation coefficient (ICC) study was conducted for both spinal curves in all tasks to ensure the repeatability and reliability of results is within the acceptable range. Shrout and Fleiss [205] call into question some past agreement of the ICC value above 0.75 shows excellent agreement between measured variables. Remarkably, the ICC values showed excellent reliability for all tasks as shown in Table 4.7. For Forward Bending and Pick Object, the ranges of ICC values are between 0.993 to 0.999 and 0.974 to 0.996, respectively. For the Weightlifting tasks, the values range between 0.941 to 0.998 and 0.947 to 0.997 for 5kg and 10kg loads, respectively.

4.4.1.2 One and Two-way ANOVA

One-way ANOVA analysis was conducted to highlight the significant kinematic difference of the thoracic and lumbar multi-segment when compared with and without the global measurement. This analysis has opted based on similar studies on spinal segmental kinematic by [33, 223]. To highlight the interaction between approaches used, Two-way ANOVA analysis was used. It is important to note that the significant value in these studies that tended to be closer to the significant level limit which is 0.05, is then assumed as significant for all thoracic and lumbar segments.

For the statistical analysis conducted using one-way ANOVA for Forward Bending between thoracic and lumbar multi-segment, the results are contrasting especially when compared between spinal regions. The significant differences ($P\text{-value} < 0.05$) were found within thoracic segments for all approaches both cycles, and also when compared with global thoracic kyphosis as shown in Table 4.8. The result is diverse within the lumbar spine as some of the segments did not display significant differences at a specific percentage of completion. During the Begin cycle, all approaches displayed a significant difference in all phases except at 10% and 40% for all approaches and only at 50% for FBSH (also when compared with global mean). For the Return cycle, no significant difference was observed in the lumbar middle phases only for FBSH and FBSB tasks. No significant differences were also observed for 60% and 80% of motion completion, for all Forward Bending approaches.

During Pick Object, the numbers of significant differences of thoracic segments when compared with and without global measurement are higher in the Return cycle, compared with the Begin. No significant difference during the Begin cycle was found only at 60% for POST and 90 to 100% for POSS as in Table 4.9. For the Return cycle, only POSS showed no significant difference, specifically at 0% of motion completion. No significant differences within the lumbar spine were more prominent during the Begin cycle and were found at different percentages of motion completion when compared between approaches. During this cycle, the initial transition (0%) for all three approaches displayed a statistically significant difference, which is also similar during the Return cycle. Similarly, the statistically significant differences in the Return cycle were found at a specific percentage of motion

completion, revealing the complex kinematic of multi-segmental contribution during the Pick Object task.

Inexplicably, the statistical analysis of both weightlifting tasks revealed approximately identical findings for both cycles as shown in Table 4.10 and 4.11. For the 5kg task, the statistical differences of the thoracic region were found for all approaches in both cycles. No significant difference was found for the 10kg task only for WL10ST, specifically at 20% and 30% of motion completion. For the lumbar region of both tasks, statistically, significant differences were found at the initial and end of Begin and Return cycles, respectively. This finding revealed that the lumbar segments provide different segmental contributions when the posture is close to the quiet standing posture. In the Begin cycle, more significant differences were found during stoop, compared with semi-squat and squat. The squat approaches used in both tasks showed the least lumbar kinematic differences in both cycles.

The two-way ANOVA analysis was conducted to understand whether there is an interaction between spinal segments and different approaches used in completing the tasks, as tabulated in Table 4.12. The most striking result to emerge from the analysis is that all tasks showed statistically different findings for thoracic segments in both cycles. This finding showed that each thoracic segment provides a different spinal segmental contribution in completing critical daily activities. Thus, this region should not be treated as the “Cinderella” region of the spine with only attention given to the cervical and lumbar[224]. The interaction within the lumbar region was found for Weightlift 5kg load during the Begin cycle and in both cycles during the Pick Object task.

Table 4. 7 Intraclass correlation coefficient values for the critical activities studied.

	Segment	Task											
		FBHH	FBSH	FBSB	POST	POSS	POSQ	WL5ST	WL5SS	WL5SQ	WL10ST	WL10SS	WL10SQ
Thoracic	T1-4	0.993 (0.989, 0.997)	0.997 (0.995, 0.999)	0.996 (0.994, 0.998)	0.991 (0.985, 0.996)	0.996 (0.992, 0.998)	0.981 (0.966, 0.991)	0.995 (0.991, 0.997)	0.995 (0.990, 0.998)	0.969 (0.940, 0.985)	0.992 (0.986, 0.996)	0.994 (0.989, 0.997)	0.964 (0.932, 0.984)
	T4-6	0.996 (0.994, 0.998)	0.999 (0.998,1. 000)	0.995 (0.989, 0.999)	0.992 (0.986, 0.996)	0.996 (0.994, 0.998)	0.983 (0.969, 0.992)	0.997 (0.994, 0.998)	0.994 (0.988, 0.997)	0.972 (0.948, 0.987)	0.994 (0.990, 0.997)	0.994 (0.989, 0.997)	0.969 (0.943, 0.986)
	T6-8	0.998 (0.996, 0.999)	0.999 (0.998,1. 000)	0.999 (0.997, 1.000)	0.993 (0.989, 0.997)	0.996 (0.992, 0.998)	0.982 (0.967, 0.991)	0.998 (0.997, 0.999)	0.992 (0.986, 0.996)	0.968 (0.941, 0.985)	0.996 (0.992, 0.998)	0.993 (0.987, 0.997)	0.966 (0.937, 0.984)
	T8-10	0.999 (0.998, 0.999)	0.998 (0.996,0. 999)	0.999 (0.998, 1.000)	0.994 (0.991, 0.997)	0.994 (0.989, 0.997)	0.976 (0.958, 0.989)	0.998 (0.997, 0.999)	0.990 (0.983, 0.995)	0.953 (0.918, 0.978)	0.996 (0.994, 0.998)	0.990 (0.983, 0.995)	0.948 (0.909, 0.975)
	Global T1-10	0.9980 (996,0. 999)	0.998 (0.997,0. 999)	0.998 (0.997, 1.000)	0.994 (0.990, 0.997)	0.996 (0.993, 0.998)	0.980 (0.964, 0.991)	0.998 (0.997, 0.999)	0.994 (0.988, 0.997)	0.965 (0.936, 0.983)	0.996 (0.993, 0.998)	0.993 (0.987, 0.997)	0.960 (0.926, 0.981)
Lumbar	T10-L1	0.999 (0.998, 0.999)	0.997 (0.995,0. 999)	0.999 (0.998, 0.999)	0.996 (0.992, 0.998)	0.993 (0.988, 0.997)	0.974 (0.954, 0.987)	0.998 (0.997, 0.999)	0.992 (0.986, 0.996)	0.941 (0.897, 0.971)	0.997 (0.994, 0.998)	0.991 (0.985, 0.996)	0.947 (0.908, 0.974)
	L1-3	0.998 (0.997, 0.999)	0.997 (0.994,0. 998)	0.998 (0.997, 0.999)	0.996 (0.993, 0.998)	0.994 (0.989, 0.997)	0.979 (0.967, 0.990)	0.998 (0.996, 0.999)	0.992 (0.986, 0.996)	0.962 (0.932, 0.982)	0.996 (0.993, 0.998)	0.990 (0.982, 0.995)	0.967 (0.941, 0.984)
	L3-5	0.997 (0.995, 0.999)	0.996 (0.993,0. 998)	0.998 (0.996, 0.999)	0.996 (0.993, 0.998)	0.993 (0.988, 0.997)	0.979 (0.963, 0.990)	0.997 (0.995, 0.999)	0.990 (0.983, 0.996)	0.975 (0.955, 0.988)	0.995 (0.991, 0.998)	0.986 (0.974, 0.994)	0.976 (0.957, 0.989)
	Global T10-L5	0.999 (0.998, 0.999)	0.998 (0.996,0. 999)	0.999 (0.998, 0.999)	0.996 (0.994, 0.998)	0.994 (0.989, 0.997)	0.978 (0.962, 0.990)	0.998 (0.997, 0.999)	0.993 (0.988, 0.997)	0.962 (0.933, 0.982)	0.997 (0.994, 0.998)	0.991 (0.984, 0.996)	0.965 (0.938, 0.983)

Table 4. 8 P-values of one-way ANOVA for different postural approaches during Forward Bending

		Begin cycle						Return cycle					
Approach		FBHH		FBSH		FBSB		FBHH		FBSH		FBSB	
Without, W/O or With, W global mean		W/O	W	W/O	W	W/O	W	W/O	W	W/O	W	W/O	W
Thoracic	0%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	10%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	20%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	30%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	40%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	50%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	60%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	70%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	80%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	90%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	100%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
Lumbar	0%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	10%	0.12	0.14	0.18	0.26	0.26	0.36	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	20%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	30%	<0.05	<0.05	0.02	0.04	0.04	0.06	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	40%	0.31	0.44	0.56	0.73	0.54	0.71	<0.05	<0.05	0.04	0.08	<0.05	<0.05
	50%	<0.05	<0.05	0.11	0.19	0.01	0.02	0.03	0.06	0.67	0.83	0.21	0.34
	60%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	0.44	0.59	0.09	0.17	0.38	0.54
	70%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	0.01	0.02	<0.05	<0.05	<0.05	<0.05
	80%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	0.82	0.86	0.49	0.63	0.83	0.89
	90%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	100%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05

Table 4.9 P-values of one-way ANOVA for different postural approaches during Pick Object.

		Begin cycle						Return cycle						
Approach		POST		POSS		POSQ		POST		POSS		POSQ		
Without, W/O or With, W global mean		W/O	W	W/O	W	W/O	W	W/O	W	W/O	W	W/O	W	
Thoracic	Percentage of motion completion	0%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	0.071	0.099	<0.05	<0.05	
		10%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	0.034	0.047	<0.05	<0.05	
		20%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	0.001	0.001	<0.05	<0.05	
		30%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	
		40%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	0.017	0.019	<0.05	<0.05	<0.05	
		50%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	
		60%	0.976	0.994	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	
		70%	<0.05	<0.05	0.001	0.001	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	
		80%	<0.05	<0.05	0.025	0.036	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	
		90%	<0.05	<0.05	0.067	0.093	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	
		100%	<0.05	<0.05	0.073	0.101	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	
Lumbar	Percentage of motion completion	0%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	0.001	0.001	0.001	
		10%	0.765	0.903	0.375	0.547	0.087	0.138	<0.05	<0.05	<0.05	0.001	0.002	0.004
		20%	0.290	0.471	0.012	0.021	0.011	0.018	0.048	0.034	<0.05	0.001	0.022	0.049
		30%	0.551	0.734	0.181	0.276	0.001	0.001	<0.05	<0.05	<0.05	<0.05	0.242	0.398
		40%	0.857	0.953	0.675	0.836	0.057	0.100	<0.05	<0.05	<0.05	0.001	0.841	0.937
		50%	0.239	0.404	0.083	0.158	0.561	0.739	0.003	0.006	0.028	0.052	0.627	0.775
		60%	0.064	0.128	0.008	0.017	0.973	0.996	0.515	0.689	0.578	0.738	0.065	0.108
		70%	0.077	0.131	0.001	0.003	0.670	0.842	0.019	0.035	<0.05	<0.05	0.004	0.007
		80%	0.552	0.705	<0.05	0.001	0.155	0.276	0.002	0.002	<0.05	<0.05	0.020	0.034
		90%	0.150	0.257	<0.05	0.001	0.012	0.028	0.828	0.913	0.936	0.969	0.649	0.802
		100%	0.148	0.259	<0.05	0.001	0.001	0.003	0.036	0.045	0.013	0.019	0.018	0.028

Table 4. 10 P-values of one-way ANOVA for different postural approaches during Weightlift 5kg load.

		Begin cycle						Return cycle					
Approach		WL5ST		WL5SS		WL5SQ		WL5ST		WL5SS		WL5SQ	
Without, W/O or With, W global mean		W/O	W	W/O	W	W/O	W	W/O	W	W/O	W	W/O	W
Thoracic	0%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	10%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	20%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	30%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	40%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	50%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	60%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	70%	0.014	0.007	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	80%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	90%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	100%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
Lumbar	0%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	0.022	0.045	0.635	0.771
	10%	0.064	0.109	0.152	0.257	0.034	0.064	<0.05	<0.05	0.021	0.044	0.714	0.827
	20%	0.064	0.098	0.013	0.025	0.377	0.528	<0.05	<0.05	0.028	0.057	0.836	0.897
	30%	0.033	0.062	0.053	0.093	0.007	0.012	<0.05	<0.05	0.051	0.097	0.696	0.797
	40%	0.883	0.963	0.827	0.934	0.046	0.082	<0.05	<0.05	0.158	0.268	0.211	0.317
	50%	0.186	0.315	0.561	0.744	0.190	0.316	0.118	0.197	0.792	0.919	0.014	0.026
	60%	0.002	0.004	0.151	0.257	0.497	0.678	0.540	0.711	0.421	0.599	<0.05	<0.05
	70%	<0.05	<0.05	0.044	0.082	0.846	0.931	<0.05	<0.05	0.004	0.007	<0.05	<0.05
	80%	<0.05	<0.05	0.021	0.042	0.925	0.961	0.559	0.707	0.381	0.525	0.007	0.008
	90%	<0.05	<0.05	0.019	0.038	0.807	0.896	0.011	0.016	0.030	0.043	0.044	0.054
	100%	<0.05	<0.05	0.019	0.038	0.711	0.827	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05

Table 4. 11 P-values of one-way ANOVA for different postural approaches during Weightlift 10kg load.

		Begin cycle						Return cycle					
Approach		WL10ST		WL10SS		WL10SQ		WL10ST		WL10SS		WL10SQ	
Without, W/O or With, W global mean		W/O	W	W/O	W	W/O	W	W/O	W	W/O	W	W/O	W
Thoracic	Percentage of motion completion												
	0%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	10%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	0.001	0.001	<0.05	<0.05	<0.05	<0.05
	20%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	0.059	0.089	<0.05	<0.05	<0.05	<0.05
	30%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	0.193	0.215	<0.05	<0.05	<0.05	<0.05
	40%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	50%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	60%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	70%	0.027	0.022	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	80%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
	90%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05
100%	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	
Lumbar	Percentage of motion completion												
	0%	<0.05	<0.05	0.001	0.002	<0.05	0.001	0.002	0.005	0.023	0.047	0.423	0.591
	10%	0.158	0.236	0.281	0.405	0.318	0.478	0.001	0.003	0.016	0.032	0.462	0.625
	20%	0.273	0.407	0.011	0.020	0.009	0.010	0.001	0.002	0.014	0.029	0.632	0.774
	30%	0.273	0.419	0.325	0.486	0.005	0.009	0.001	0.003	0.021	0.043	0.890	0.944
	40%	0.761	0.897	0.916	0.978	0.089	0.162	0.014	0.031	0.072	0.133	0.912	0.956
	50%	0.098	0.175	0.562	0.748	0.416	0.603	0.296	0.464	0.661	0.827	0.377	0.550
	60%	0.001	0.001	0.278	0.441	0.788	0.908	0.926	0.977	0.391	0.572	0.024	0.041
	70%	<0.05	<0.05	0.133	0.237	0.969	0.988	0.030	0.051	0.012	0.021	0.001	0.001
	80%	<0.05	<0.05	0.070	0.134	0.833	0.925	0.793	0.853	0.771	0.862	0.149	0.190
	90%	<0.05	<0.05	0.036	0.071	0.559	0.719	0.005	0.008	0.009	0.012	0.002	0.003
100%	<0.05	<0.05	0.023	0.047	0.388	0.548	<0.05	<0.05	<0.05	<0.05	<0.05	<0.05	

Table 4. 12 P-values of two-way ANOVA for ROM of different forward bending approaches.

		Cycle			
		Begin		Return	
	Task \ Region	Thoracic	Lumbar	Thoracic	Lumbar
Interaction	FB	<0.05	0.989	<0.05	0.976
	PO	<0.05	<0.05	<0.05	<0.05
	WL5	<0.05	0.023	<0.05	0.117
	WL10	<0.05	0.256	<0.05	0.450

4.4.2 Discussion

4.4.2.1 Comparison of the data with the literature

Similar literature studies that explored the segmental kinematic of the thoracic and lumbar regions used different boundaries for both of the spinal regions. As mentioned in subsection 3.4.2, the limit between the thoracic and lumbar region in this study was set on the 10th, not the 12th thoracic vertebra as commonly used by most literature. This study chose this vertebra as the boundary between the thoracic and lumbar region on the basis of variation in facet joint orientation [215] and radiographs of standing posture[31]. Thus, the angle discrepancy between this study and the literature is to be expected. A study by Ignasiak, Rüeger [38] that study on the thoracic and lumbar multi-segmental thoracic spine during maximum forward bending which is similar to the current study, also implemented a different angle normalization method. Furthermore, the critical daily activities in this study used a variety of different approaches for each task, lessening the resemblance between thoracic and lumbar segmental kinematics with the literature studies. Moreover, the study implements different spinal segmental endpoints for both spinal regions with other literature as mentioned previously. Comparing the global thoracic and lumbar angles when standing is the best way to ensure that the findings in this study are consistent with other literature. This is due to the fact that much of the literature, especially for dynamic tasks, identifies the starting point while standing. Tables 4.13 and 4.14 show the comparison for both spinal regions. These values correlate fairly well with similar literature that utilize non-radiography studies on healthy subjects and further support the data reliability of current study.

Table 4. 13 Comparison of global thoracic kyphosis with similar studies during standing.

	Forward Bending			Pick Object			Weightlift 5kg load			Weightlift 10kg load		
	FBHH	FBSH	FBSB	POST	POSS	POSQ	WL5ST	WL5SS	WL5SQ	WL10ST	WL10SS	WL10SQ
This study	35.43± 5.21	35.58± 4.99	34.32± 5.10	37.09± 9.60	36.15± 6.62	34.71± 5.71	36.69± 5.44	36.43± 6.16	35.58± 5.63	37.03± 6.26	36.45± 5.48	35.50± 5.89
Kuo, Tully [17], Kuo, Tully [36]	38.30 ± 8.20											
Tully, Fotoohabadi [108]	31.10 ± 8.70											
Hey, Teo [225]	33.67 ± 10.45											

Table 4. 14 Comparison of global lumbar lordosis with similar studies during standing.

	Forward Bending			Pick Object			Weightlift 5kg load			Weightlift 10kg load		
	FBHH	FBSH	FBSB	POST	POSS	POSQ	WL5ST	WL5SS	WL5SQ	WL10ST	WL10SS	WL10SQ
This study	19.05± 5.69	19.56± 6.48	19.34± 6.04	20.17 ± 6.92	18.78± 7.41	18.44± 6.82	18.98± 7.33	19.28± 7.24	18.62± 7.72	18.78± 7.52	18.31± 7.41	19.33± 7.60
Kuo, Tully [17], Kuo, Tully [36]	16.00 ± 5.60											
Ippersiel, Robbins [133]	20.46±6.07											
Tully, Fotoohabadi [108]	16.20 ± 8.30											
Shum, Crosbie [216]	14.00 ± 2.00											

4.4.2.2 Forward Bending

At the early stage of forward bending transition, the upper thoracic segment, T1-8 displayed a noteworthy main difference and contribution compared with the lower thoracic segment, T8-10 (Figure 4.7: a), specifically within 0-25% phases. During forward bending in a standing posture, the sacral acts as a pivot to allow trunk flexion. The farther the spinal segment from the sacral, the greater the movement thus suggesting the reason for the high increment of upper thoracic segments compared with the lower segment. Critically, approximately between 40 to 50% of motion completion, all thoracic segments and global thoracic reach peak flexion with approximately 30° of angular displacement increments (Figure 4.8: a). These findings offer vital evidence that suggests peak mobility of the thoracic spine during forward bending occurs at the middle transition for all segments. After exiting this phase, the upper thoracic segment, T1-4, revealed solely an extension curve for FBSB and FBHH until maximal forward bending was reached. By limiting the range of motion, forward bending with FBSH reduces mobility and minimizes thoracic segmental extension. This finding provides further evidence for correct postural implementation during forward bending by suggesting hand support at the knee. It is also important to note that the extension of the T1-4 segment is greater when no support is introduced as in Figure 4.7(a). At the 60% phase, all approaches share similarities with high increment at the last thoracic segment, T8-10, but smaller in magnitude for SH. Interestingly, T1-4 began to contribute less till the end of the first cycle within the same phase. At the end of the Begin cycle, forward bending with FBSB and FBHH shows similar characteristics with extension at T1-8 segments. Interestingly, minimal change in angular displacement is displayed during FBSH compared with other approaches. Also, the maximization of hand extension at the knee resulted in a delay for T1-4 segments.

During the Return cycle, minimal thoracic segmental mobility was observed at a 20% phase for FBSH compared with FBSB and FBHH (Figure 4.7: b). At 50 and 60% phases, all segments tend to have a peak increment of extension with the T8-10 segment displaying higher magnitude compared with other segments for all approaches. In the Return cycle, the difference in thoracic segmental contribution between approaches is significant, especially between 0 and 40% of motion

completion. From 50 percent until the end of the forward bending task, all three techniques showed nearly identical thoracic segmental increment (Figure 4.8: a). This discovery is noteworthy since the alternative strategy used during forward bending only has an impact on the first 40% of the task completion. These results offer vital evidence for the correlation between forward bending approaches and the extension increment pattern during the Return cycle. Also, at 70%, each thoracic segment displayed a high extension increment, but lower than the previous phase thus suggesting both phases play a crucial point for spinal stabilization as the changes of the CG take place. During forward bending, the CG of a human body is closer to the ground compared to quiet standing. The single most conspicuous result to emerge from this is that only the lowest thoracic segment, T8-10 demonstrated flexion increment at the end of the Return cycle. This finding is observable only if the spine is being studied segmentally compared by assuming the thoracic spine only as a single segment.

It is crucial to note that there is no significant difference and interaction between forward bending approaches to lumbar ROM. The change in angular displacements between different approaches was also almost identical for the Begin and Return cycles. These values correlate favorably with Alessa and Ning [226] and further support the concept of minimal lumbar contribution during forward bending. In their analysis, lumbar and abdominal muscle behaviors were substantially reduced and indicated that a change of loading from lumbar active tissues to passive tissues occurs spontaneously during the execution of short-term, static trunk bending tasks.

During the Begin cycle of the lumbar region, there is minimal segmental contribution between 0-20% phases for all approaches as in Figure 4.8 (b). High flexion increment is noticeable within 40 to 60% phases with the middle phase recording the highest increment. This implies that spinal stabilization occurs within these phases as the CG tends to shift downward as the trunk flexes. As suggested by du Rose, Breen [227], the activation of the lumbar segmental muscle that acts as the damping mechanism decreases irregular displacements during bending and resulted in limited angular displacement changes. Comparing segments, top lumbar segments, T10-L1, showed greater mobility compared with the other segments. This is understandable as the segment is anatomically farther than other lumbar segments

from the sacral. Also, it is important to note that the lowest lumbar segment, L3-5 displayed a lower flexion increment. When shown as in Figure 4.7 (c) and (d), the difference in the lumbar segmental contribution is insignificant since the contribution is small. Thus, the normalized segmental difference plot, as shown in Figure 4.8, is the ideal approach to analyze the changes in lumbar segmental kinematics.

For the Return cycle within the lumbar, the segmental contribution at 20% is approximately similar for all segments. Comparable with the Begin cycle, the middle phase of the Return cycle showed a peak increment of the extension curve up to 30° for both the segmental and global measurement. It is also noticeable that at 60%, the increment toward extension is considerably high but lower than the previous phase, as the Begin forward bending cycle. Inexplicably, forward bending without (FBHH) and minimal (FBSB) hand support, does result in a higher change in angular displacement of the lumbar spine compared with maximum utilization of the hand flexion at the knee, especially in the Begin cycle. This discovery is noteworthy since the alternative strategy used during forward bending only has an impact on the first 40% of the task completion. This finding offers compelling evidence for right postural implementation to reduce excessive movement of the back spine by using a hand as a support at the knee. Similar to the thoracic, all three approaches displayed approximately identical lumbar segmental increments starting 10% until the end of the forward bending task. This is an intriguing discovery since the differences in postural implementation during forward bending only had a substantial impact on the Begin cycle, not the Return cycle. Using the global measurement to understand lumbar kinematic is oversimplified, thus neglecting the actual segmental contribution even though the change is minimal.

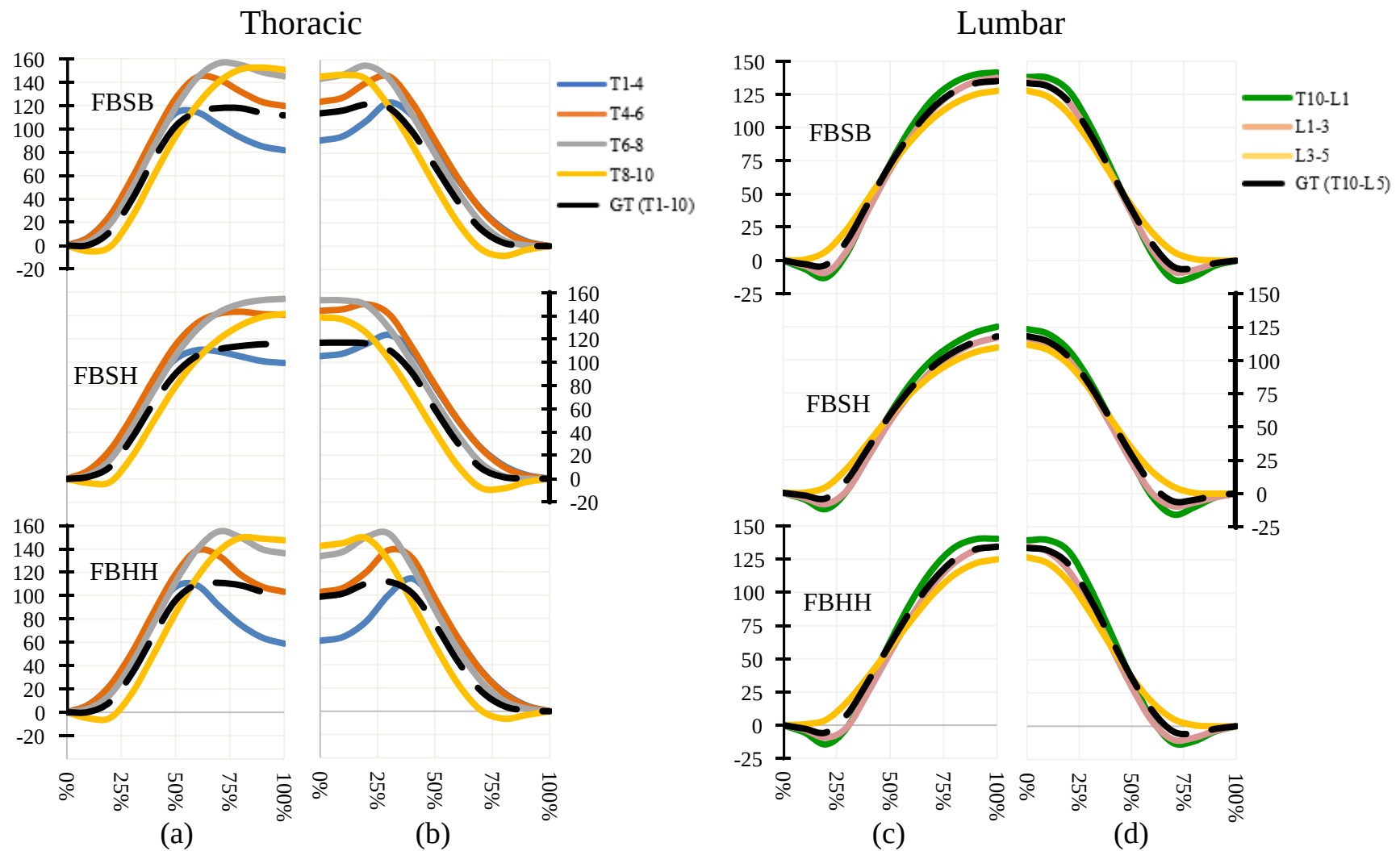


Figure 4. 7 Normalized thoracic and lumbar Mean difference during Forward Bending for Begin (a&c) and Return (b&d) cycles.

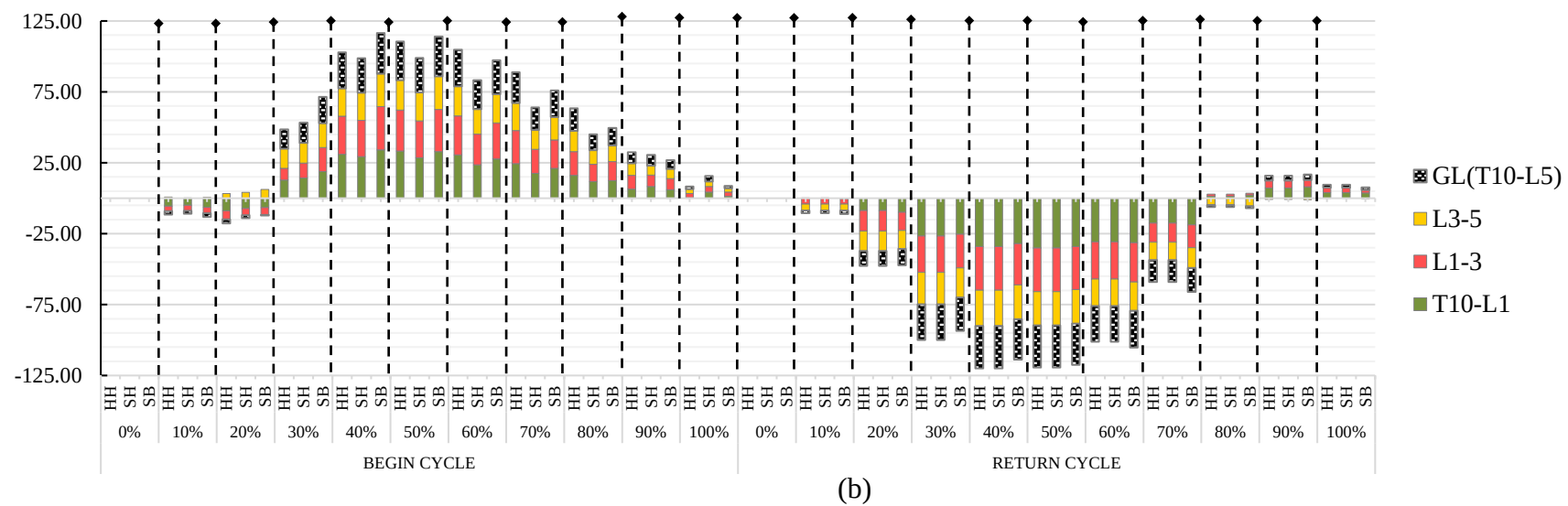
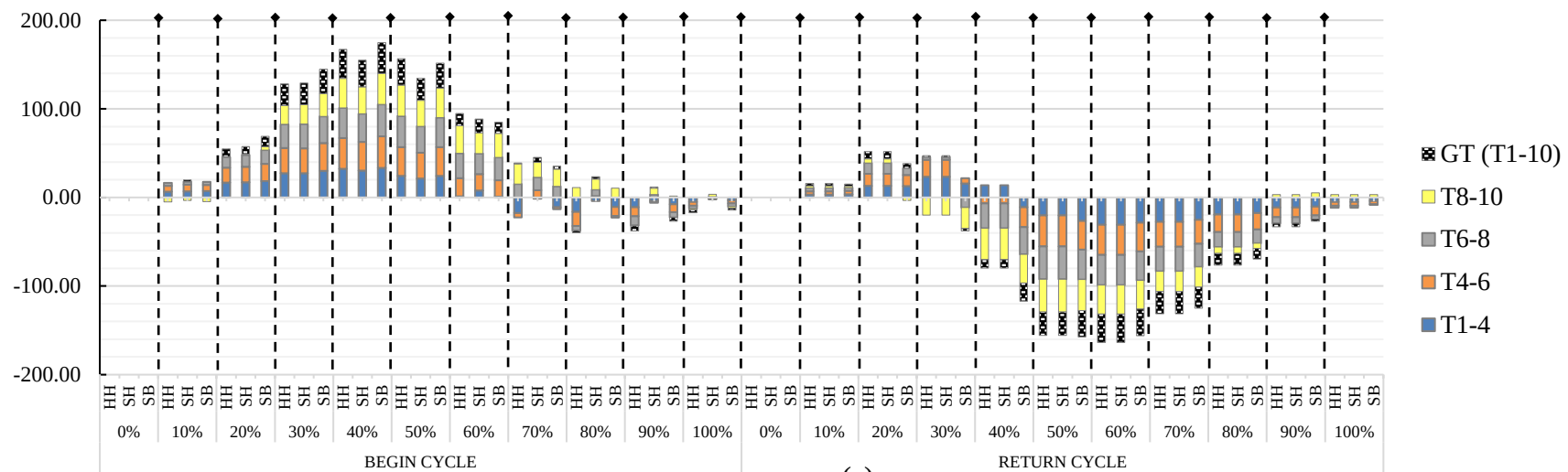


Figure 4. 8 Normalized Segmental Difference during Forward Bending for Thoracic (a) and Lumbar (b).

4.4.2.3 Pick an Object on the Floor

As previously stated, subjects were instructed to pick an object with a small mass from the floor in this task. This study only focused on the effects of different postural approaches on the spine, thus the effect of the object's mass was neglected. Different people implement different approaches based on their needs. However, most of the time, this choice comes with negligence of the consequences of bad posture implementation, especially at the lower back. This study compared the difference of segmental kinematics of the thoracic and lumbar spine during three different pick object approaches, which are stoop, semi-squat, and squat.

Diversity of the mean difference of thoracic segments is more obvious during POST particularly after the middle phase in the Begin cycle, compared with POSS and POSQ (Figure 4.9: a). Similarly, the Return cycle (Figure 4.9: b) also showed POST with large discrepancies of the segmental kinematic. Interestingly, the change of mean difference within POSS and POSQ occurs steadily with a delay after a 50% phase. This delay takes place quicker with POSQ compared with POSS. It also resulted in a minimal change in the thoracic segmental kinematic as in Figure 4.10. Comparing these three approaches, the squat method displayed the lowest peak flexion for each thoracic segment compared with POST and POSS. From the statistical study conducted, the global thoracic kyphosis measurement is significantly different for all three approaches and there are minimal numbers of no significant findings for POST and POSS. This result has further strengthened this research's hypothesis that none of the spinal segments share similar kinematic with global measurement. For the Begin cycle, the peak flexion increment of the thoracic segment was reached at the approximately middle transition for all approaches with POST displaying greater increment as displayed in Figure 4.9. Leaving this phase, the stoop approach displayed a chaotic segmental pattern with all thoracic segments tending to be extended. Within the same phase, the kinematic tended to be different when the subject implemented the semi-squat and squat approach as they showed minimal flexion increment with no segmental extension. Extensive mobility of the spinal segment does relate to the high usage of spinal muscle at the back that consequently promotes back pain. As discussed by Panjabi [228], the change of the muscle recruitment strategy is required during

complex body movements with the temporary intention of maximizing spine stability beyond normal requirements. This suggests that high segmental discrepancy can lead to back pain. Large intersegmental changes may result in an increased demand for passive spine structures. Minimal change in the thoracic kinematic was observed for POSS and POSQ at the final phase of the Begin cycle compared with POST, which showed an increment of segmental extension. Comparing POSS and POSQ, the latter approach was shown to be more preferable with no drastic change within the segmental angle, thus suggesting the best approach toward picking an object is by implementing a squat posture.

During the Return cycle of the thoracic region, approximately minimal or no change of angular displacement was observed within POSS and POSQ approach between 0 to 30% phases as in Figure 4.10. During this phase, minimal or no spinal mobility is needed (only for Semi-squat and Squat) as the change of momentum in the vertical direction requires an extension of the knee, thus reducing the back muscle's excessive activity within the thoracic spine. However, flexion increment was observable during POST at the same phase, indicating poor posture implementation compared with the two approaches earlier. Comparatively, POST when picking up objects showed similar excessive segmental increments toward extension during the Return cycle, reaching up to 40°. Notably, it occurred at 60 to 70% phases just before the subject completely standing from bending forward. This drastic change is not the same for POSS and POSQ approach that displays extension increments only up to 25° and 15°, respectively.

As the lumbar region is anatomically closer to the sacral that acts as a pivot for the trunk flexion, it is expected to have a smaller change in segmental kinematic compared to the thoracic region. Figure 4.9 (c) and (d) show that the mean differences for both cycles are less than those for the thoracic area, as predicted. For the Begin cycle, there is a spike of lumbar flexion increment, obviously within the upper lumbar segment during POST, compared with POSS and POSQ Figure 4.11. The latter approaches displayed smaller changes with POSQ showing the most nominal increment. It is important to note that at the initial of the Begin cycle (0-20% phases), all lumbar segments displayed passive kinematic for all approaches. Moreover, this

passive kinematic is also evident at the end phases (90-100%), especially for POSS and POSQ approaches, thus highlighting controlled lumbar segmental and minimal muscle movement.

The result within the initial phases (0-25%) of the Return cycle of all approaches showed minimal lumbar segmental increment (Figure 4.11). As the subject tries to stand up from picking an object up from the floor, the body will stabilize itself by coordinating the CG closer to the top of the feet. During this coordination, the spinal motion, especially within the lumbar spine, should be minimal thus suggesting the reason for this kinematic. However, it is important to highlight the flexion that occurred during this phase, which occurred only during POST. The curve pattern between the Begin and Return cycles is approximately identical for all approaches. The significant change in angular displacement, notably during POST, demonstrates the difference between the approaches. The most surprising finding is the POST resulted in 3.5 times greater angular displacement while POSS only 2.5 when compared with the POSQ approach (Figure 4.9: c and d). All lumbar segments displayed approximately similar contributions during pick objects as the difference between segments is not prominent ($P\text{-value} > 0.05$) especially in the midst of the transition. In summary, during this task, lumbar segments only tend to be active or have higher mobility between 30 to 70% for both cycles. Also, the implementation of POSS and POSQ approaches has successfully minimized and controlled excessive movement of lumbar multi segments compared with POST, which can promote lesser load on the lumbar back muscle.

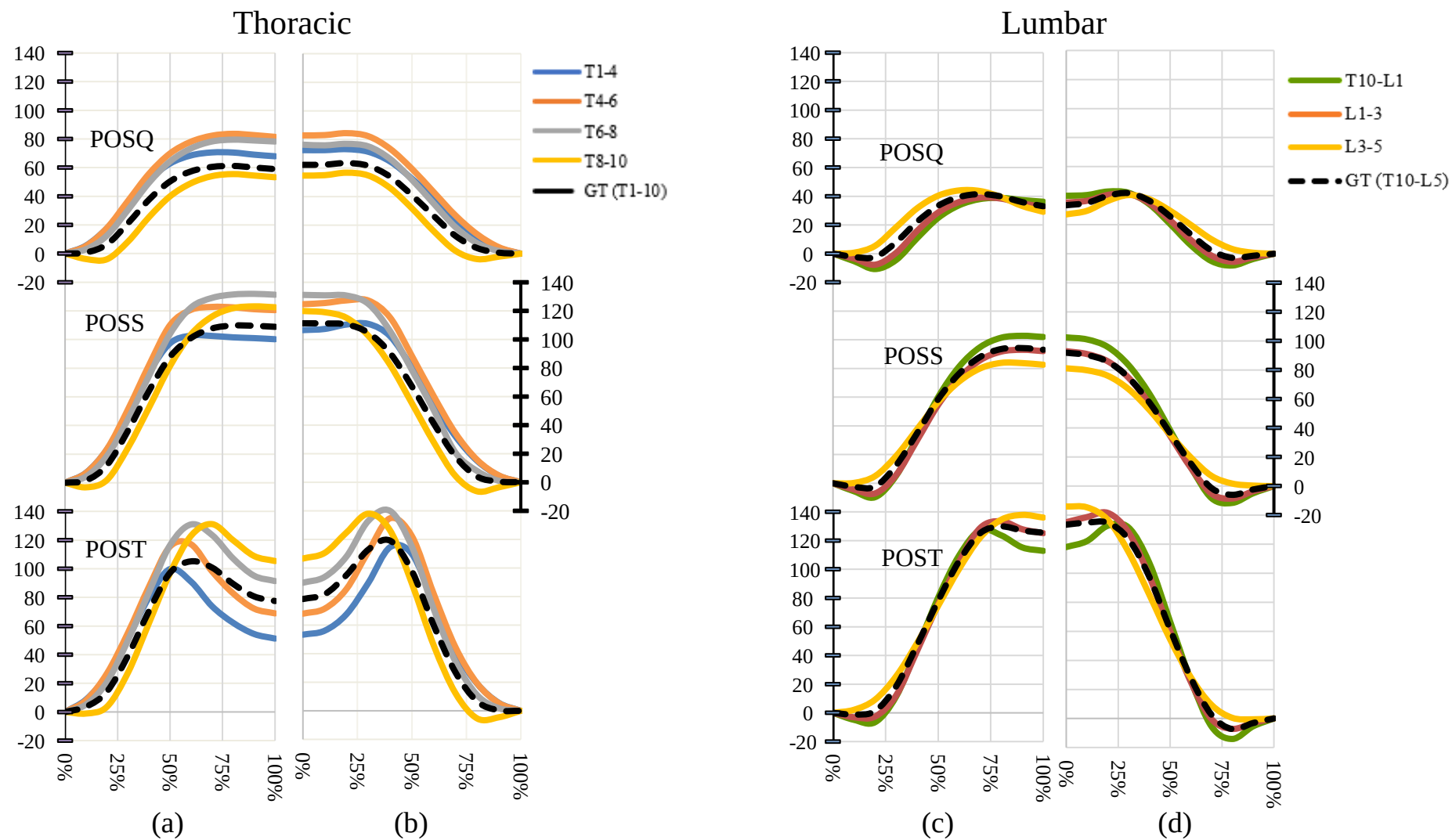


Figure 4. 9 Normalized thoracic and lumbar Mean Difference during Pick Object for Begin (a&c) and Return (b&d) cycles.

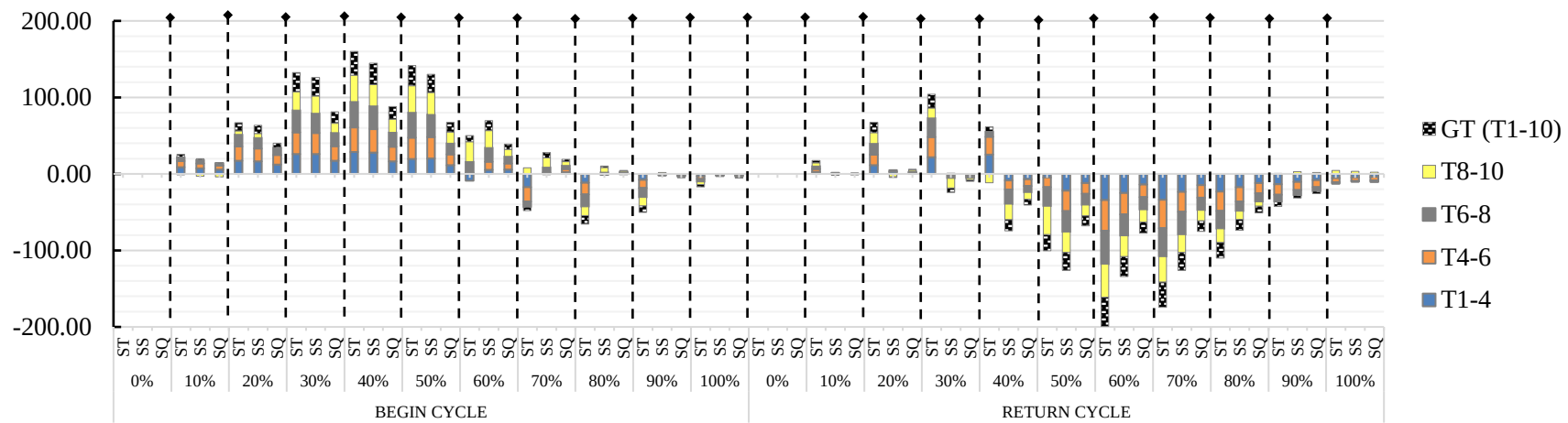


Figure 4.10 Normalized thoracic Segmental Difference during Pick Object

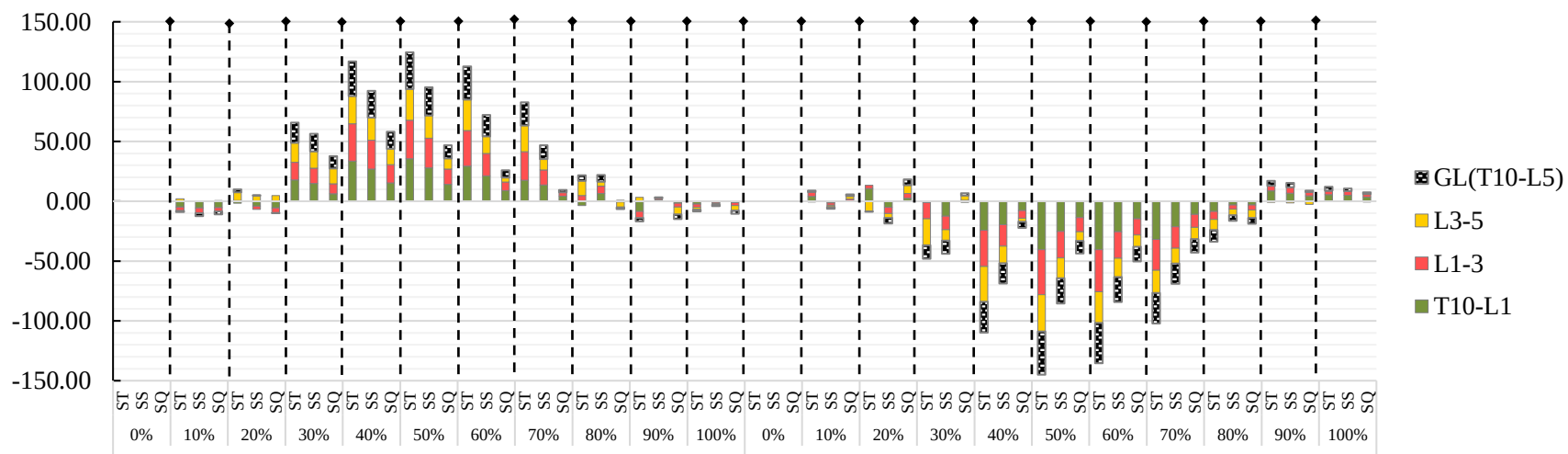


Figure 4.11 Normalized lumbar Segmental Difference during Pick Object

4.4.2.4 Weightlifting 5kg and 10kg loads

On average, weightlift with stoop posture showed the highest mean difference within the thoracic region for both load cases especially amid the transition as in Figures 4.12 (c, and e) compared with other approaches. The restriction flexion at the knee resulted in the need for excessive trunk flexion when performing WL5ST and WL10ST. The other two approaches implement knee flexion thus reducing the dependency for flexion at the spine as the subject reached the load on the floor, resulting in lesser increment toward thoracic flexion. Moreover, both load cases in the stoop approach showed larger angular displacement differences between 30 to 60% phases of the Begin cycle, similar to the pick objects pattern. During stoop, the top thoracic segment (T1-6) displayed a contrasting change in the kinematic approach toward extension while other segments tend to flex more. There is a good probability that even though the trunk continues to flex forward, unconsciously the head tends to extend backward at a slower pace. In Beaudette, Briar [229] investigation into this, it was found that ROM of specific spine regions during forward-flexion is affected by head and gaze orientations. They stated that the head postures requiring an extension of the neck may also facilitate extension within the thoracic spine (i.e. C7-T1 to T4-5) thus supporting the finding of the present study. Critically, during WL5ST and WL10ST, the lowest thoracic segments, T8-10, displayed the highest flexion compared with other approaches, especially between 40 and 70% phases, thus indicating excessive segmental maneuverability. Concurrently, this can lead to unnecessary usage of muscle energy at the back of the spine. Within the same phase and segment, the WL5SQ and WL10SQ approaches only showed minimal thoracic segmental flexion increment. At the early phases (0-20%) for both loading conditions and all approaches, T8-10 displayed the least segmental increment, thus highlighting the passive kinematic of this segment (Figures 4.14: a, and b). This indicates the acceptability level of assuming this segment as the thoracic segmental endpoint as it is anatomically located near the lumbar spine, which behaves similarly. Interestingly, all three approaches showed a distant change of angular kinematics after the middle of the weightlifting task (Figure 4.12). The high segmental disparity was observed for stoop while semi-squat and squat approaches showed a slower change of segmental

kinematic, especially for the latter approach. This concurs well with Mörl, Bradl [230] as the squat approach resulted in minimal segmental change during weightlifting.

During the next cycle in which the subjects try to lift the load in the vertical direction, the difference of the thoracic kinematics for all three approaches is evident. WL5ST and WL10ST showed a higher extension increment which is approximately two times higher than WL5SQ and WL10SQ (Figures 4.12: b, and d). This finding does highlight the need for corrective postural implementation in daily tasks to avoid factors that can lead to back pain in the long run. Semi-squat and squat approaches have successfully proven to result in the minimum requirement of thoracic segmental mobility, thus reducing unnecessary efforts on the muscle of the back, especially during load-carrying tasks. It is interesting to note that between the end of the Begin and initial Return cycles, there is a delay in the angular displacements of the thoracic segments for all approaches, which is more prominent for both semi-squat and squat approaches. All segments tend to display passive kinematics with minimal increment starting from the 80% phase until the end of Begin cycle. The results point to the likelihood of an effective load-sharing condition between multi-segment as the subject starts to carry the load. In contrast, the stoop approach displayed a variant in segmental increment with an increase in segmental extension as going down the thoracic segments.

Interestingly, both loading conditions showed identical lumbar segmental kinematics for all approaches with no significant difference ($P\text{-value} > 0.05$), thus highlighting that the range of load used does not affect the lumbar segments. The most striking result to emerge from the data is that the peak of angular displacement achieved by implementing the stoop approach is approximately 4 times higher than the squat for both loads carried, hence emphasizing the importance to carry out weightlifting tasks with proper posture. Higher angular displacement during flexion will result in a greater moment that needs to be countered by the erector spinae muscle at the back which can promote LBP. At the initial transition specifically from 0 to 20% phases, all lumbar segments showed the least contribution in all approaches. Two phases that displayed significant changes in the Begin cycle are between 30 to 60% phases as they showed a more evident change of segmental kinematic (Figures 4.15:

a, and b). The top lumbar segment, T10-L1, showed the highest increment compared with other segments at these phases. The results point to the probability that high loads and changes in the vertebral kinematics in the spine do not generally occur as usually predicted at full upper body flexion but moderate angles of flexion[231]. This study also revealed the contribution of both global or localized lumbar segments is minimal at the end of the initial task with the knee flexed (semi-squat and squat) compared with a stoop. There is a strong probability that weightlifting without knee flexed (stoop) increase the need for maximum spinal flexion which can result in excessive muscle usage for stabilization purpose. When comparing the highest to lowest change in segmental kinematics, stoop approaches comes out on top for both cycles, whereas squat comes in last.

Interestingly, at the initial phase of the Return cycle for the lumbar region, only the squat approach showed progress toward minimal flexion while the other two were going for extension (Figures 4.13: b, and d). This may result due to the displacement of the subject body's CG in the forward to vertical directions that only occurred during the Squat. This approach showed excellent control of spinal flexion by making use of maximum knee flexion as subjects try to stand from the squat posture. Also, the further the distance of the load from the body's CG, especially with the addition of load, the greater the extension curve. This would also appear to indicate that the controlled spinal and trunk flexion during squat do affect the kinematic direction of spinal segments. Similar to the previous cycle, between 30 to 70% phase, the changes of segmental differences were evident for all approaches (Figures 4.15: a, and b). All lumbar segments showed approximately equal segmental contribution toward extension with the upper segment (T10-L1) displaying the highest increment. The remarkable result to emerge from the data is that stoop approaches displayed the highest segmental increment toward an extension, compare with other approaches. Some preliminary work was carried out several years ago by Mitnitski, Yahia [232] which suggests that the nervous system controls muscle forces to restrict curvature at the lumbar level, which restricts the displacement of vertebrae when the load increases. This would safeguard the loaded lumbar spine from excessive compressive forces in a flexion position thus suggesting the occurrence of minimal difference between lumbar segments. As previously discussed, WL5ST and WL10ST also showed the highest

angular displacement shift amongst the approaches used in the Return cycle. It is critical to note that the upper lumbar segment (T10-L1) tends to be highly active as the subject reaches complete quiet standing for both cycles. In contrast, other segments demonstrated passive kinematics, including global measurement. This study has successfully demonstrated that the squat approach is capable of reducing intersegmental movements [230] and is preferable during frequent or heavy lifting.

4.5 Active and passive spinal segments.

One of the main objectives of this research is to identify the active-passive spinal segments for common daily tasks studied. However, it was no easy task as none of the literature has defined it, especially the limit of the segments that have lower or higher mobility. As previously discussed, the kinematics of spinal segments within the thoracic and lumbar were complex, even in the same task while being approached differently. Understanding this kinematic requires depth analysis as the spine has 6 DOF, which makes the analysis becoming challenging. The DOF allows the motion of flexion, extension, left and right lateral bending, and left and right rotation of the spine. The 6 DOF is made up of three translational and three rotational motions, along with three axes, each of which has a quantifiable stiffness. The spine bends left and right around a coronal plane axis, rotates left and right around a transverse plane axis, and flexes and extends along a sagittal plane axis as in Figure 2.3. Alternatively, this study tries to come with a new method in defining the boundary of the active and passive segment during the quick change of spinal kinematics. The active segment is described in this study as a segment with significant motion, whereas the passive segment has low mobility.

It is essential to refocus the attention on the reason for controlling the spinal segment. The kinematics of the spine is affected in individuals who suffer from LBP, as discussed in detail in the Literature Review. The indication of LBP is related to the muscle activation that supports the back of the body, especially within the spine or known as erector spinal muscle. Controlling muscle activation thus means controlling LBP's impact. As the current study did not investigate the electromyography (EMG)

analysis of the back muscles, this study relies on a resemblance with similar studies, that utilizes the trunk angle. Some researchers tried to explore the muscle activation at the trunk and back muscles in completing specific tasks[227, 233-240]. However, the research that has the similarity between the current study is few in numbers. As discussed, the current study includes forward bending, pick an object on the floor, and weightlifting tasks with different approaches. Takahashi and Kikuchi's research on the EMG activities during forward bending and weightlifting activities is very comparable to the current study, particularly in terms of the demographic profile of the participants and the experimental procedures[241]. The details of the comparison are shown in Table 4.13 below.

Table 4. 15 Comparison between the other researcher and current study.

	Takahashi and Kikuchi	Current study
Similarities	Forward bend without external weight.	Forward bend with hanging hand.
	Forward bend with an external weight of 10kg.	Weightlift 10kg load with stoop posture.
	Average subjects' age: 25 BMI: 23.37	Average subjects' age: 24.67±4.56 BMI: 22.99±3.78
Difference	Their study only evaluated up to 30 degrees of forward flexion, whereas the current study addresses maximal flexion.	

The EMG reading at the erector spinal muscle (ESM) was greatest during forward bending with a 10kg load, especially when the trunk angle was at 30°, according to their findings. In the same way, the ESM reading without the load was the highest as shown in Figure 4.16. Considering the maximum EMG activation, the angle of the trunk at 30° is used as the critical limit for the current study. For the current analysis, the trunk angle at 30° is approximately at 40% of motion completion for all tasks as shown in Figures 4.17 and 4.18. Thus, the value of angular displacement at this phase for both thoracic and lumbar multi-segment will be taken as the maximum value. The calculation will only utilize 50% assistive support because the goal of the proposed multi-joint spinal orthosis is to assist the user rather than fully support the spinal kinematics. Thus, the maximum value of the angular displacement will be multiplied by 0.50 to get the active-passive limit. For example, the maximum

segmental value of top thoracic segment T1-4 during FBHH is 32.27°. The value is then multiplied by 0.50 to get the active-passive limit of T1-4 during FBHH, which is 16.13°. The same calculation is applied for WL10ST too. Due to limited data on a similar study, thus the study was only able to apply the calculation for both of these tasks. The calculation for the other tasks will be the main focus of future work. The tabulated limit values for both of the tasks are shown in Table 4.14 below.

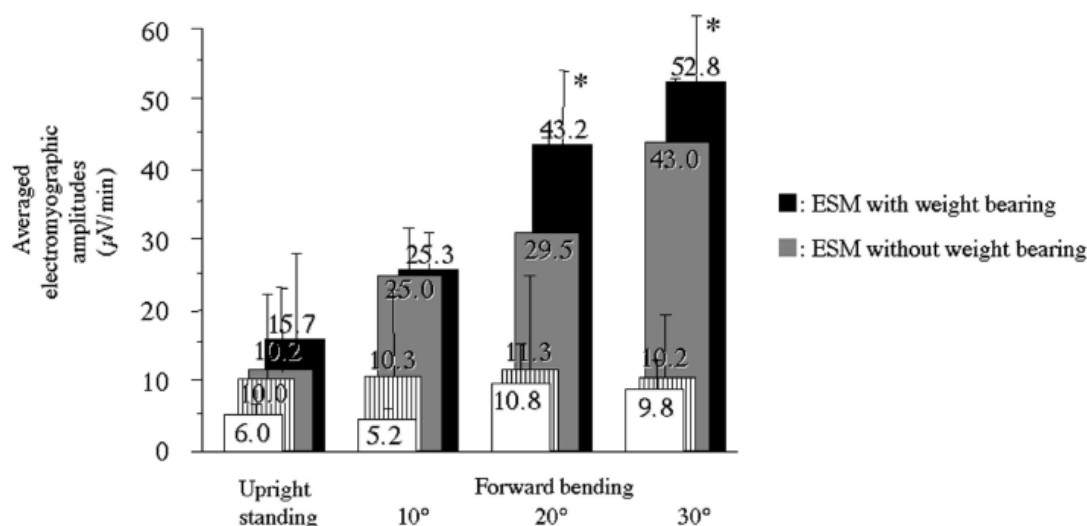


Figure 4. 12 Relationship between the load of the lumbar spine and activities of the trunk muscles during upright standing and bending forward 10°, 20°, and 30°, respectively. Source: [239]

Table 4. 16 Active-Passive limits during FBHH and WL10ST.

Task	Thoracic				Lumbar		
	T1-4	T4-6	T6-8	T8-10	T10-L1	L1-3	L3-5
FBHH	16.13	17.30	17.13	16.72	15.54	13.40	9.75
WL10ST	39.73	43.50	39.68	26.45	17.53	16.55	20.90

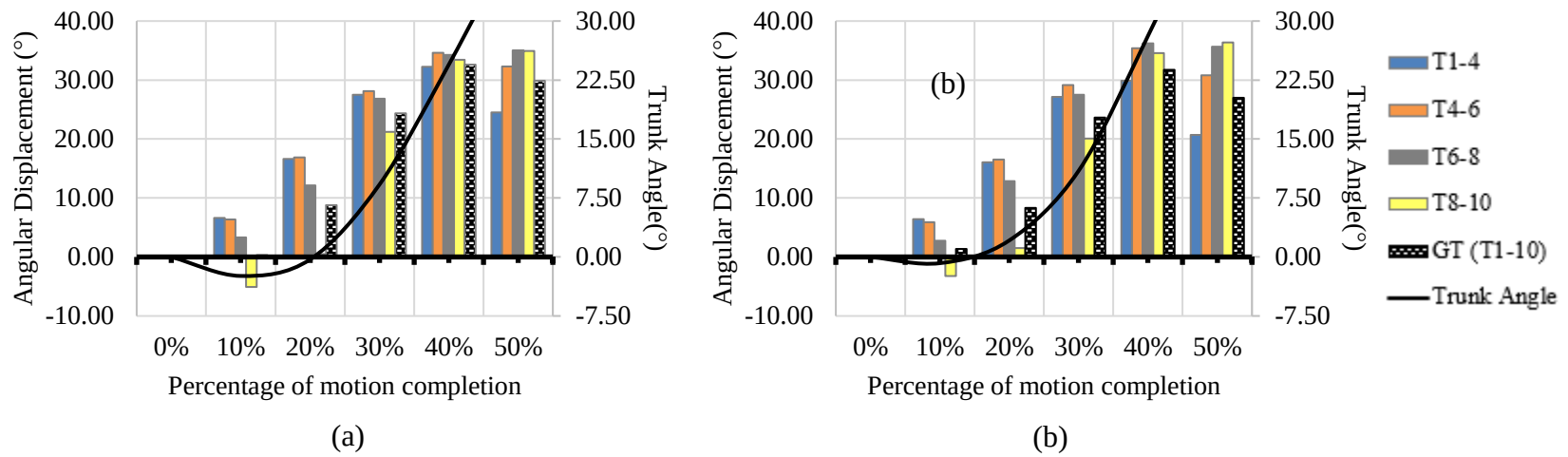


Figure 4.13 Thoracic multi-segment and Trunk angle for (a) FBHH and (b) WL10ST

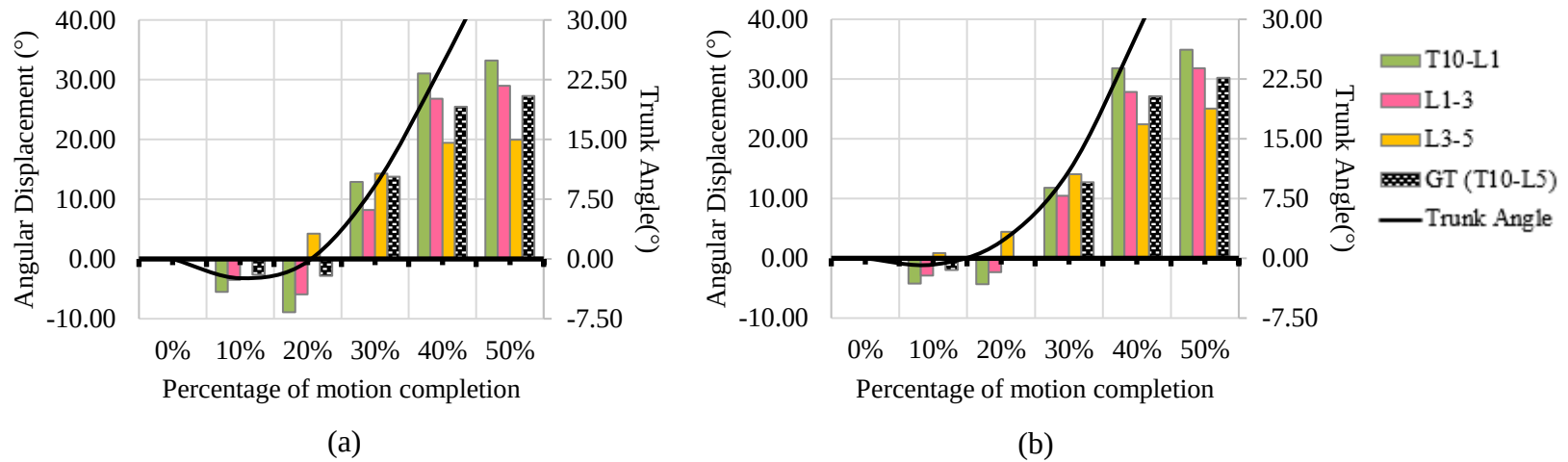


Figure 4.14 Lumbar multi-segment and Trunk angle for (a) FBHH and (b) WL10ST

4.6 Contribution of hip and knee in daily activities

In completing most of the daily tasks, the contribution of hip and knee angles is undeniably important. Every task requires particular ranges of hip and knee contributions which make them unique in every way. As this study includes common daily tasks, the contribution of these angles is unquestionably important. This section will discuss the variation of hip and knee angle Mean and Segmental contribution towards the completion of the tasks studied.

4.6.1 Results

The results of normalized means and segmental differences of hip and knee angles were shown in the tables based on different approaches for Forward Bending (Table 4.77), pick an object on the floor (Table 4.78), and weightlifting for both load cases (Table 4.79-80). Using this data, the combo chart was used to represent the findings in a simpler form. The statistical One-way ANOVA revealed no significant difference for Mean and Segmental differences were found for the hip angles in all tasks, even between both cycles. For the knee angle, the statistical findings were distinctive. During forward bending, only the Mean difference showed a statistically significant difference when compared between different approaches (P-value: 0.002[Begin] and 0.013[Return]). However, that is not the case during the Pick an object and Weightlift tasks. The knee angles were significantly different only for Segmental during Pick Object and Weightlift 5kg load in both cycles. Due to similar approaches used for Pick and Object and Weightlift tasks, one-way ANOVA was also conducted to compare the difference of these tasks as in Table 4.76.

4.6.2 Discussion

The data presents the difference in the kinematics of hip and knee angles studied during common daily tasks. For forward bending, only the SB approach resulted in flexion in the knee while others towards extension in the Begin cycle as in Figure 4.22 (a, c, and e). The knee kinematics behaved in reverse for the Return cycle (Figure 4.22: b, d, and f). Broadly speaking, the knee kinematics and contribution

during forward bending were found to be minimal. Oppositely, the hip angle displayed greater contribution with SB top the chart compared with SH and HH. The peak segmental differences of the hip were significantly higher within 50 and 75% phases for both cycles which indicates significant contribution toward forward bending occurs at these phases. Unexpectedly, comparing between cycles, the Begin cycle displayed slightly higher segmental differences than the Return only for SB and SH tasks. The absence of hands on the knee that acts as support could well be responsible for this result.

For the Pick Object and Weighlift tasks, the changes of Mean and Segmental differences for the hip angle are identicals based on different approaches used (Figure 4.23,4.24 and 4.25: a-f). Based on the statistical analysis used to compare these tasks, the results have successfully strengthened this conviction. There is no significant difference found within the hip angle. However, the case is not similar to the knee angle. The Mean Difference showed no significant difference for all approaches except for Stoop in the Begin cycle. Interestingly, all results for knee Segmental Differences displayed a statistically significant difference in both cycles. There is a strong probability that the presence of load does only affect the knee's kinematics.

Table 4. 17 P-values for Pick an object, weightlift 5, and 10kg loads.

Difference	Region	Begin Cycle			Return Cycle		
		Stoop	Semi Squat	Squat	Stoop	Semi Squat	Squat
Mean	Hip	0.884	0.934	0.982	0.846	0.900	0.900
	Knee	0.011	0.993	0.981	0.212	0.976	0.986
Segmental	Hip	0.993	0.970	0.997	0.999	0.981	0.997
	Knee	0.027	0.040	0.036	0.007	0.048	0.032

Table 4. 18 Normalize mean and segmental difference for hip and knee angle during Forward Bending

			Percentage of motion completion					P-value
Cycle	Region	Task	0%	25%	50%	75%	100%	
Mean Difference	Begin	SB	0.00	-6.18±14.70	-49.36±13.11	-76.04±10.35	-84.66±9.87	0.990
		Hip	SH	0.00	-6.18±14.70	-49.36±13.11	-76.04±10.35	
		HH	0.00	-3.49±15.40	-51.04±17.18	-84.04±15.41	-93.21±14.5	
		Knee	SB	0.00	-1.50±0.99	-6.06±3.54	-7.31±4.18	0.002
		SH	0.00	1.50±0.99	6.06±3.54	7.31±4.18	7.41±4.94	
		HH	0.00	1.57±0.96	5.79±2.78	6.20±5.07	6.15±6.08	
	Return	SB	-83.26±13.15	-69.43±16.84	-29.31±22.67	-0.74±10.85	0.00	0.998
		Hip	SH	-83.26±13.15	-69.43±16.84	-29.31±22.67	-0.74±10.85	
		HH	-91.87±15.48	-74.16±12.71	-26.74±16.65	3.76±9.71	0.00	
		Knee	SB	-8.18±5.24	-6.33±3.87	-3.25±4.69	-0.13±2.56	0.013
		SH	8.18±5.24	6.33±3.87	3.25±4.69	0.13±2.56	0.00	
		HH	6.93±8.06	5.64±5.03	2.15±3.54	0.17±2.40	0.00	
Segmental Difference	Begin	SB	0.00	-6.18±14.70	-43.18±5.59	-26.67±6.97	-8.62±3.23	0.986
		Hip	SH	0.00	-6.18±14.70	-43.18±5.59	-26.67±6.97	
		HH	0.00	-3.49±15.40	-47.55±10.63	-33.01±5.61	-9.17±1.63	
		Knee	SB	0.00	1.50±0.99	4.56±3.04	1.24±1.94	0.969
		SH	0.00	1.50±0.99	4.56±3.04	1.24±1.94	0.10±1.14	
		HH	0.00	1.57±0.96	4.22±2.38	0.41±2.92	-0.05±1.24	
	Return	SB	0.00	13.83±6.56	40.11±8.89	28.58±14.68	0.74±10.85	0.986
		Hip	SH	0.00	13.83±6.56	40.11±8.89	28.58±14.68	
		HH	0.00	17.70±6.00	47.42±8.96	30.50±14.84	-3.76±9.71	
		Knee	SB	0.00	-1.85±5.16	-3.09±3.00	-3.12±3.27	0.955
		SH	0.00	-1.85±5.16	-3.09±3.00	-3.12±3.27	-0.13±2.56	
		HH	0.00	-1.29±4.51	-3.49±3.79	-1.98±2.31	-0.17±2.40	

Table 4. 19 Normalize mean and segmental difference for hip and knee angle during pick object

			Percentage of motion completion					P-value	
Cycle	Region	Task	0%	25%	50%	75%	100%		
Mean Difference	Begin	SB	0.00	-5.93±16.93	-57.00±21.08	-99.61±15.21	-111.21±11.38	0.958	
		Hip	SH	0.00	-10.56±16.72	-70.18±25.37	-111.13±22.39		-122.40±18.91
		HH	0.00	-9.60±16.28	-55.25±42.56	-96±43.49	-110.25±41.98		
		Knee	SB	0.00	0.64±2.26	3.18±4.43	-0.44±10.20	-1.29±11.82	0.073
			SH	0.00	-9.44±6.35	-42.13±12.60	-75.79±15.36	-90.12±15.86	
			HH	0.00	-15.17±7.55	-55.64±40.37	-100.46±42.98	-117.63±43.75	
	Return	Hip	SB	-109.85±14.22	-96.49±18.72	-49.97±26.1	-3.58±15.43	0.00	0.988
			SH	-119.88±16.96	-101.26±15.41	-50.77±15.94	-4.15±14.78	0.00	
			HH	-119.15±25.33	-101.66±25.84	-55.79±24.83	-9.66±16.93	0.00	
		Knee	SB	-2.85±12.29	-1.88±10.82	-1.76±6.73	-1.35±2.13	0.00	0.082
			SH	-90±16.74	-73.75±15.13	-37.99±10.58	-8.67±4.46	0.00	
			HH	-125.56±13.82	-106.04±12.14	-60.94±11.75	-16.03±7.00	0.00	
Segmental Difference	Begin	SB	0.00	-5.93±16.93	-51.08±7.91	-42.61±9.89	-11.60±4.78	0.987	
		Hip	SH	0.00	-10.56±16.72	-58.62±11.48	-41.94±5.8		-11.27±5.66
		HH	0.00	-10.10±16.56	-52.78±15.63	-36.76±13.69	-14.32±9.88		
		Knee	SB	0.00	0.64±2.26	2.54±5.47	-3.62±6.39	-0.85±3.16	0.051
			SH	0.00	-9.44±6.35	-32.89±9.67	-33.46±9.24	-14.32±5.95	
			HH	0.00	-15.97±6.84	-48.25±9.50	-41.08±15.53	-17.40±6.65	
	Return	Hip	SB	0.00	13.35±7.49	46.52±8.98	46.38±16.88	3.58±15.43	0.988
			SH	0.00	18.62±6.27	50.49±10.11	46.61±8.29	4.15±14.78	
			HH	0.00	17.49±9.52	45.87±8.65	46.13±13.04	9.66±16.93	
		Knee	SB	0.00	0.97±2.90	0.11±7.25	0.41±5.74	1.35±2.13	0.047
			SH	0.00	16.24±4.61	35.77±8.03	29.32±7.92	8.67±4.46	
			HH	0.00	19.52±9.29	45.10±7.07	44.91±7.29	16.03±7.00	

Table 4. 20 Normalize mean and segmental difference for hip and knee angle during weightlift 5kg load

			Percentage of motion completion					P-value
Cycle	Region	Task	0%	25%	50%	75%	100%	
Mean Difference	Begin	SB	0.00	-0.95±13.89	-41.75±16.45	-76.25±11.79	-84.33±11.35	0.916
		SH	0.00	-10.08±17.47	-59.57±24.86	-90.89±19.88	-97.51±18.00	
		HH	0.00	-7.89±13.66	-54.62±24.46	-85.44±26.83	-96.47±25.58	
	Knee	SB	0.00	1.08±1.58	3.00±3.69	2.31±7.49	1.00±9.05	0.054
		SH	0.00	-12.23±8.01	-42.08±17.05	-70.14±15.63	-79.70±14.06	
		HH	0.00	-16.27±8.10	-58.4±16.81	-94.96±18.94	-111.91±17.36	
	Return	SB	-83.26±17.9	-75.06±17.73	-33.74±19.47	3.79±10.13	0.00	0.968
		SH	-92.24±19.82	-84.33±21.07	-42.01±23.49	-0.46±12.49	0.00	
		HH	-93.98±26.41	-81.53±26.86	-39.9±25.16	-1.62±14.04	0.00	
	Knee	SB	-2.71±12.96	-1.94±11.79	-0.58±8.13	-0.39±1.87	0.00	0.080
		SH	-80.82±13.11	-69.06±10.99	-35.41±8.17	-8.08±4.87	0.00	
		HH	-116.15±17.04	-94.94±13.99	-55.45±9.74	-14.33±5.75	0.00	
Segmental Difference	Begin	SB	0.00	-0.95±13.89	-40.79±8.66	-34.51±8.09	-8.08±4.61	0.965
		SH	0.00	-10.08±17.47	-49.49±9.37	-31.32±7.40	-6.62±5.44	
		HH	0.00	-7.89±13.66	-46.73±14.15	-30.81±9.37	-11.03±9.54	
	Knee	SB	0.00	1.08±1.58	1.92±3.81	-0.69±5.03	-1.31±4.85	0.035
		SH	0.00	-12.23±8.01	-29.86±10.77	-28.05±6.61	-9.56±4.77	
		HH	0.00	-16.27±8.10	-42.12±10.14	-36.56±6.3	-16.95±8.47	
	Return	SB	0.00	8.20±4.17	41.32±7.70	37.53±12.93	-3.79±10.13	0.977
		SH	0.00	10.06±5.81	42.32±6.42	41.55±13.01	0.46±12.49	
		HH	0.00	15.04±7.03	41.63±5.56	38.28±16.33	1.62±14.04	
	Knee	SB	0.00	0.78±1.49	1.36±4.17	0.19±6.66	0.39±1.87	0.050
		SH	0.00	10.89±3.96	33.64±11.94	27.34±5.30	8.08±4.87	
		HH	0.00	19.01±7.95	39.48±9.60	41.12±7.87	14.33±5.75	

Table 4. 21 Normalize mean and segmental difference for hip and knee angle during weightlift 10kg load

			Percentage of motion completion					P-value
Cycle	Region	Task	0%	25%	50%	75%	100%	
Mean Difference	Begin	SB	0.00	-9.58±19.43	-55.28±28.07	-84.36±20.02	-90.74±17.76	0.963
		SH	0.00	-16.69±21.73	-69.95±30.32	-92.41±23.84	-98.25±20.53	
		HH	0.00	-14.13±21.98	-63.67±29.16	-88.48±28.93	-97.86±27.18	
	Knee	SB	0.00	-1.09±3.96	-2.41±12.5	-5.87±19.53	-7.42±22.85	0.059
		SH	0.00	-12.92±8.62	-47.87±16.37	-71.68±18.67	-79.14±18.11	
		HH	0	-22.67±6.7	-71.35±14.39	-102.75±15.48	-114.41±17.98	
	Return	SB	-83.57±19.99	-76.24±21.39	-34.66±20.39	5.31±11.15	0.00	0.978
		SH	-94.43±22.64	-82.48±21.44	-37.75±22.61	2.70±12.76	0.00	
		HH	-93.13±26.48	-82.47±23.88	-41.63±19.96	0.97±13.35	0.00	
	Knee	SB	-4.8±19.24	-7.89±20.44	-3.6±13.2	-0.79±3.39	0.00	0.089
		SH	-80.33±16.65	-66.10±16.68	-33.12±12.85	-5.76±4.93	0.00	
		HH	-117.75±14.08	-99.55±11.04	-59.65±10.72	-15.87±6.66	0.00	
Segmental Difference	Begin	SB	0.00	-9.58±19.43	-45.70±10.54	-29.08±10.44	-6.38±3.47	0.991
		SH	0.00	-16.69±21.73	-53.26±11.1	-22.46±8.22	-5.84±5.35	
		HH	0.00	-14.13±21.98	-49.54±13.44	-24.82±8.29	-9.38±4.31	
	Knee	SB	0.00	-1.09±3.96	-1.32±9.17	-3.45±8.78	-1.55±3.88	0.071
		SH	0.00	-12.92±8.62	-34.95±9.96	-23.81±11	-7.45±5.15	
		HH	0.00	-22.67±6.7	-48.67±12.4	-31.4±10.06	-11.67±6.71	
	Return	SB	0.00	9.57±4.62	41.58±8.13	39.97±11.7	-5.31±11.15	0.989
		SH	0.00	11.94±6.93	44.73±5.37	40.45±12.97	-2.70±12.76	
		HH	0.00	12.92±5.87	40.84±6.23	42.6±12.58	-0.97±13.35	
	Knee	SB	0.00	0.85±3.43	4.29±8.79	2.81±10.82	0.79±3.39	0.069
		SH	0.00	14.23±4.16	32.99±9.60	27.36±10.34	5.76±4.93	
		HH	0.00	16.67±7.29	39.9±8.08	43.78±6.67	15.87±6.66	

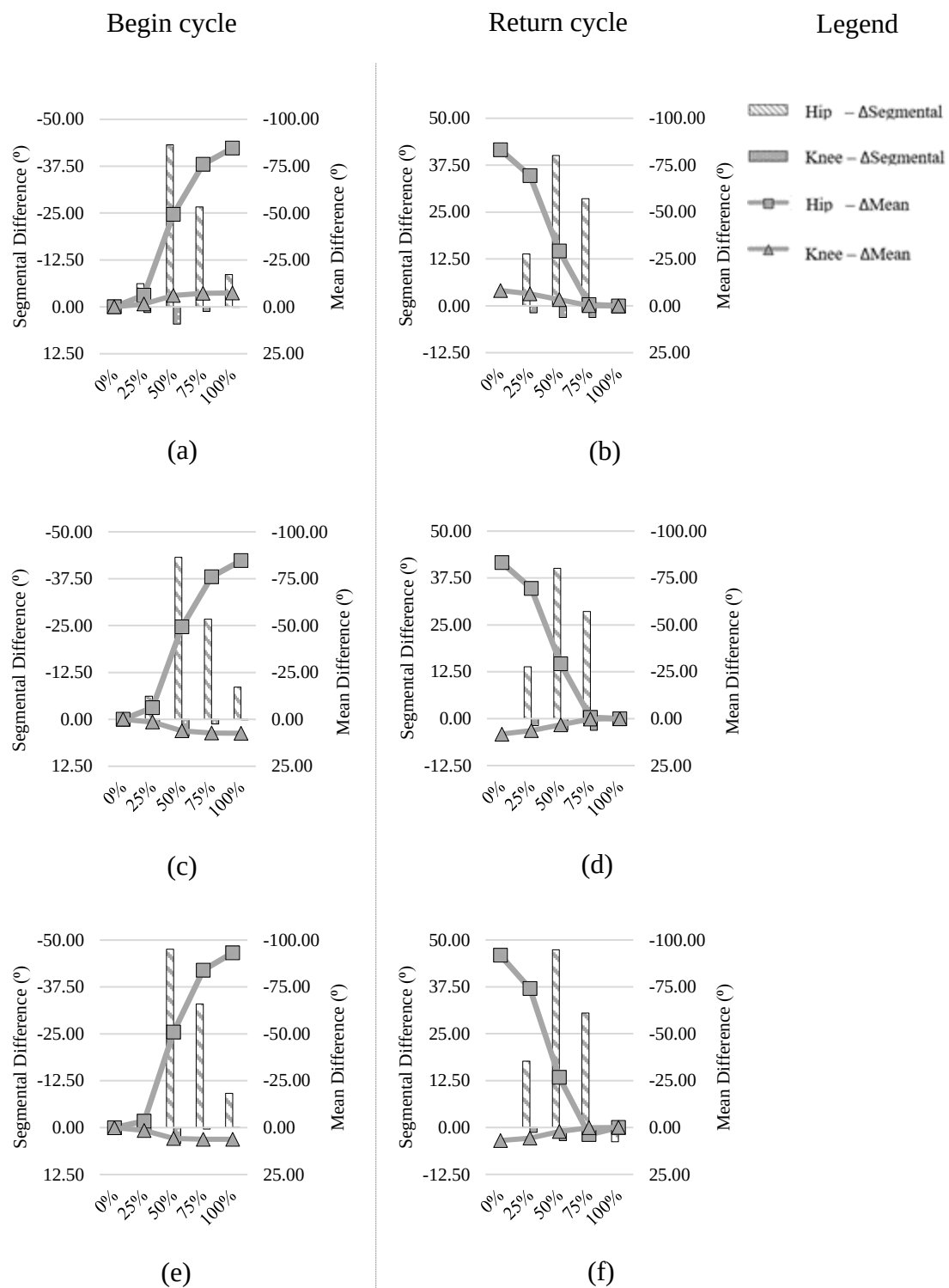


Figure 4. 15 Normalized results of hip and knee for Main and Segmental differences during Forward Bending.

*Straight Back (a & b), Straight Hand (c & d), and Hanging Hand (e & f).

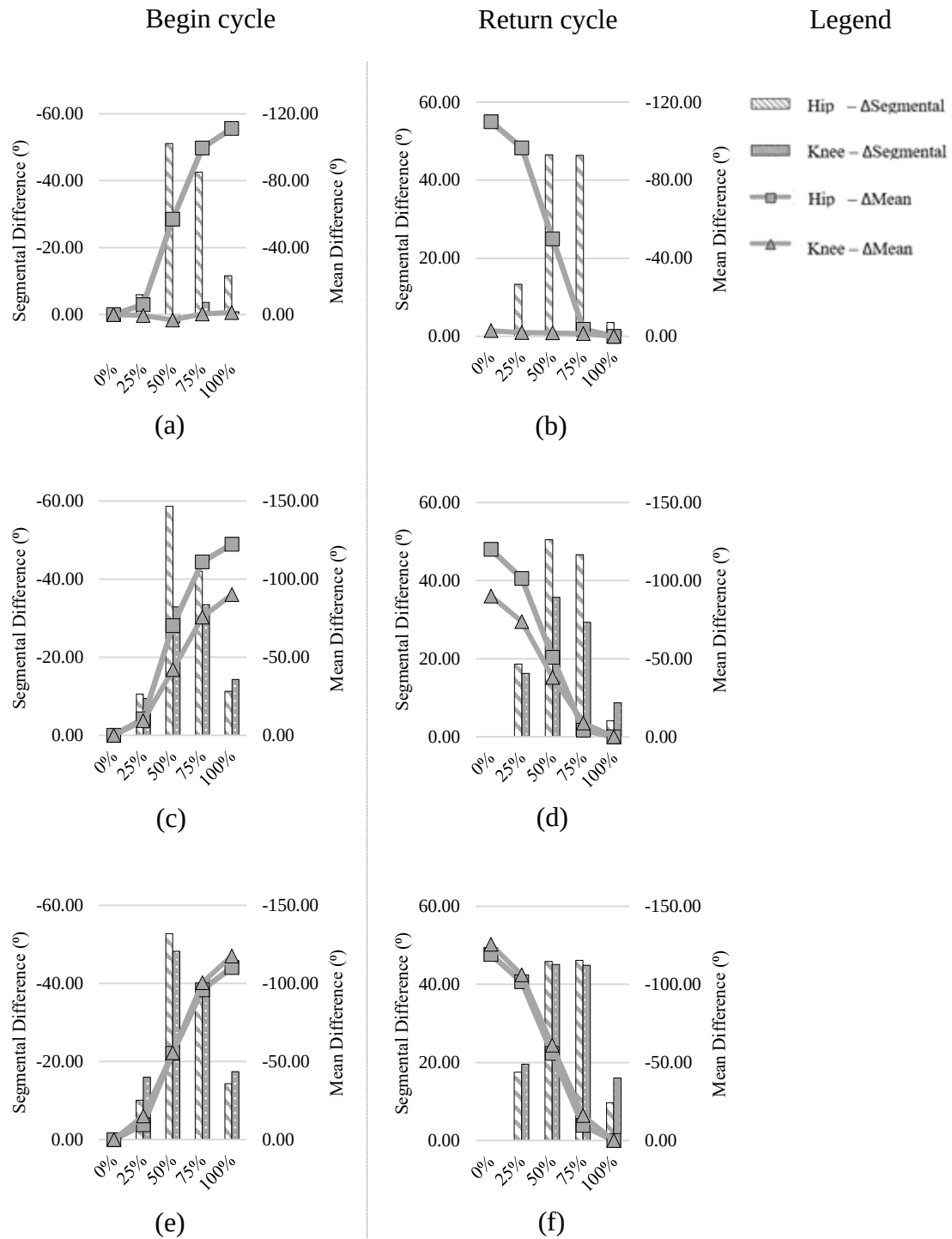


Figure 4. 16 Normalized results of hip and knee for Main and Segmental differences during pick object

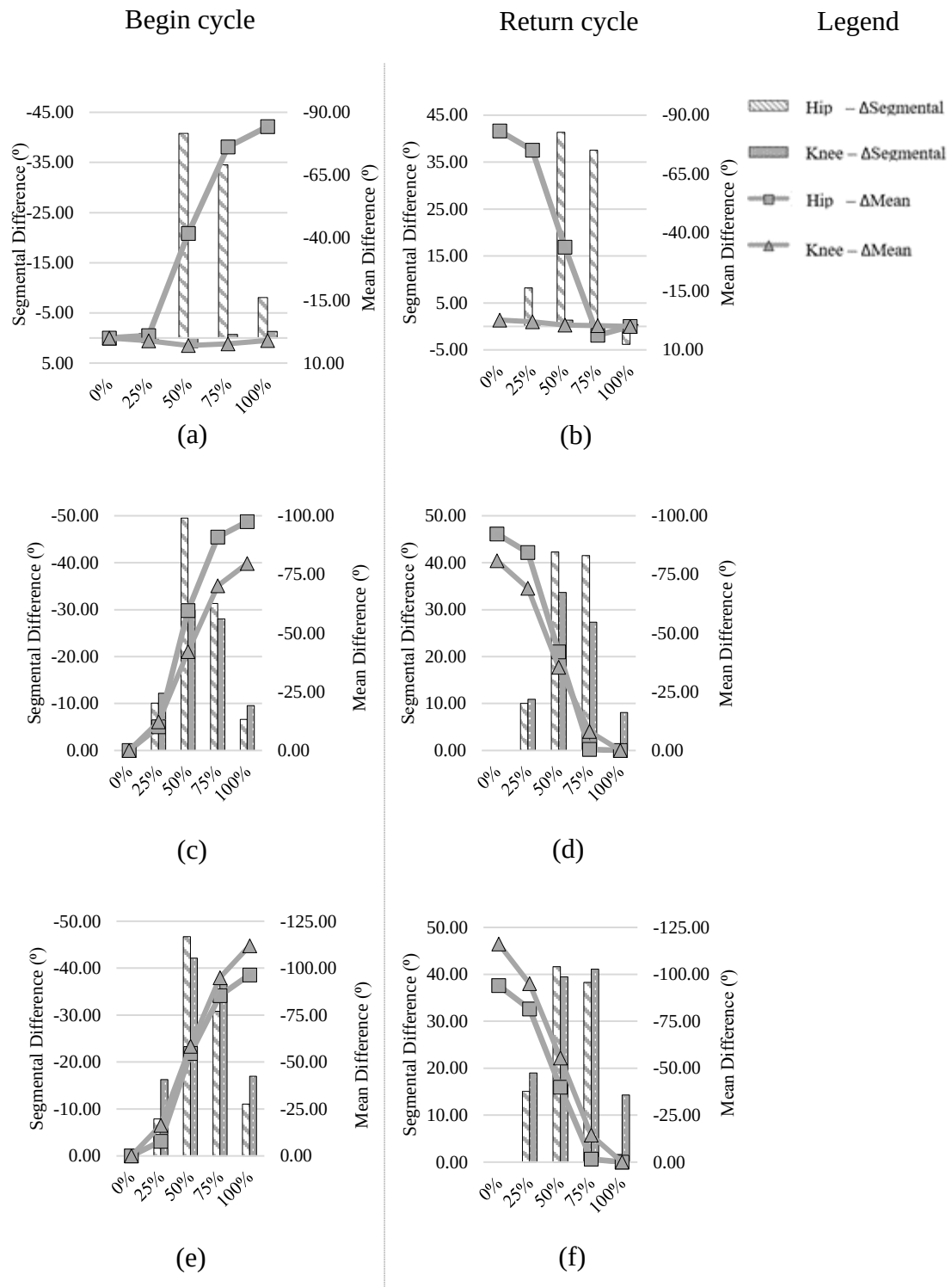


Figure 4. 17 Normalized results of hip and knee for Main and Segmental differences during Weightlift 5kg load

*Stoop (a & b), Semi Squat (c & d) and Squat (e & f).

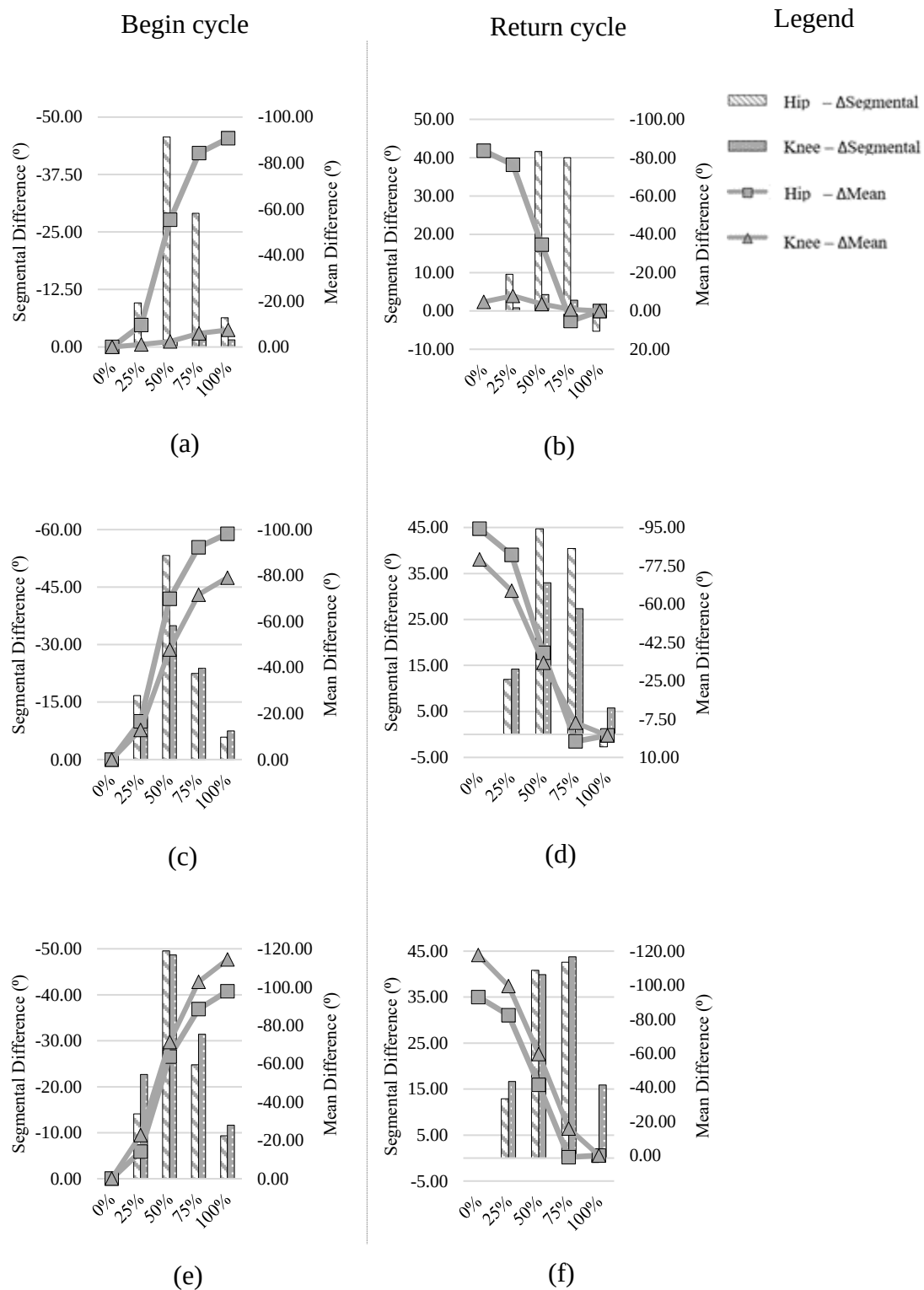


Figure 4. 18 Normalized results of hip and knee for Main and Segmental differences during Weightlift 10kg load

*Stoop (a & b), Semi Squat (c & d) and Squat (e & f).

CHAPTER 5

PROPOSED DESIGN AND CONTROL IMPLEMENTATION

5.1 Final Concept Design

In the Literature Review, the pluses and minuses of commercial and ongoing researches of passive and active spinal orthoses are discussed in detail, specifically based on robotic system types, actuators and sensors, supported movements, and control systems. For the design development, this study has completed the product design specifications (PDS), morphological chart, selection matrix, and final concept. As the PDS is a key document that defines what the product must do[242], this study took into account aspects of commercially available spinal orthosis such as quality and reliability, user interface, material, cost comparison, as well as standard and regulation (Appendix A). Then, the alternatives solutions for the multi-joint spinal orthosis were generated in form of the morphological chart based on the design from the literature as in Appendix B. This chart depicted the critical features of the proposed spinal orthosis, including the type of actuator and sensor, harness design, material for each segment, power supply, and emergency mechanism. Four initial concepts were presented using this chart as summarised in Appendix C. The hand-drawn sketches for the suggested joint mechanism are in Appendix D.

A two-stage concept selection approach was used to select the best concept, with the first step focusing on concept screening and the second stage on concept scoring. The concept screening was conducted using a selection matrix to compare each concept against Spinomed, a commercially available and flexible spinal orthosis (Appendix E). This orthosis was chosen because it has an excellent rehabilitative design, covers all three curves of the spine from cervical to lumbar, and has received a lot of favorable feedback from users[165, 168, 169]. The proposed multi-joint spinal orthosis in this study is designed to support and self-control different segments of the

human spine's three flexible curves based on the needs of the individual. Thus, the aspect of flexibility is crucial. The first and final concepts were seen to be preferable from the reference, therefore they were given more consideration. The second and third concepts were discarded owing to production issues and design complexity, as well as the use of unsuitable material. The second stage, which is the concept scoring, was used since it is a more thorough form of concept screening. The rubric table was used as a guideline on how to rate every four concepts presented, as shown in Appendix F. This stage took into account five essential criteria: ease of handling, maintenance, ease of manufacturing, smoothness, safety measures, and form design as in Appendix G. Each criterion was allocated a weighting percentage depending on its cruciality. Ease of Handling (30%), Maintenance (20%), and Safety (20%) are three factors that received a high percentage. The Ease of Manufacturing, Smoothness of Operation, and Aesthetic Factor each received a 10% percentile value. The rating was given for each criterion, which resulted in a Weightage score(W/s), based on the weighted value of the criteria. Based on the score, the first (3.72), third (3.62), and final (3.60) concepts were considered for the final design concept due to the high weightage score.

The final concept design of a multi-joint spinal orthosis integrated with a simple, yet efficient control system is proposed based on findings from the literature and design development results. This concept design is the combination between the first and third concept proposed, based on the results of the concept screening and concept scoring as in Appendix H. The maximum reachable angle of motion for the vertebra is 25° (the first cervical vertebra). Using this value, the initial locking mechanism for a minimum of two vertebrae is designed. The mechanical system for the proposed design of the dynamic multi-joint spinal orthosis consists of 8 components, which are the base, locking system, locking system cover, and screw, rod, locking rod holder, and connector rods. The main components of the system are shown in Figures 5.1, 5.2, 5.3, and 5.4.

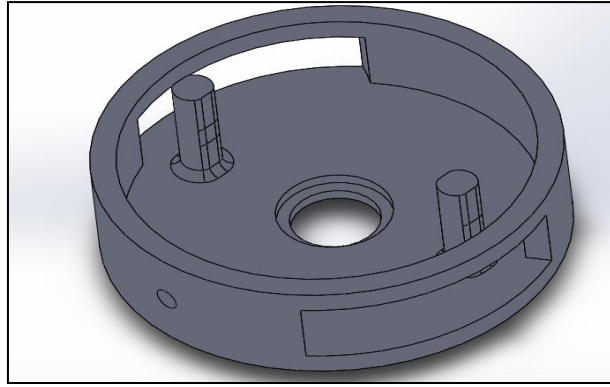


Figure 5. 1 Locking system

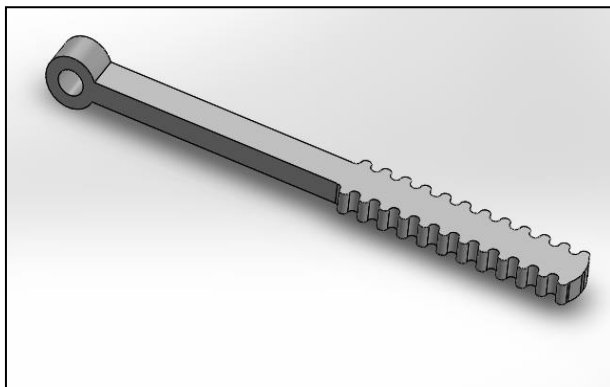


Figure 5. 2 Rack

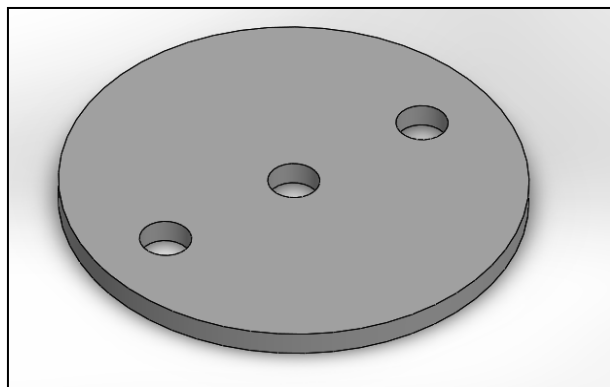


Figure 5. 3 Locking system cover

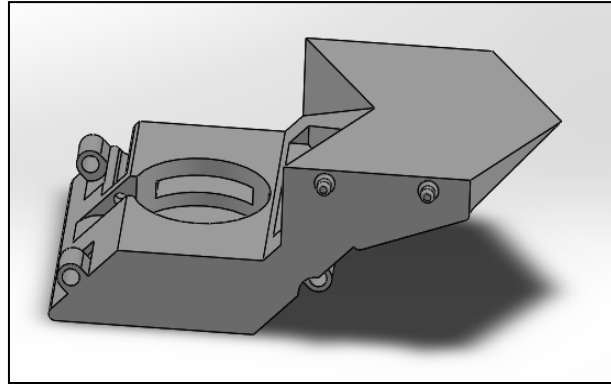


Figure 5. 4 Base

The dimension of the proposed spinal orthosis is made almost similar to the size of two human vertebrae. Thus, each segment should be able to control the motion of flexion-extension for a minimum of 2 vertebrae simultaneously depending on the location of the spinal curves specified. For a recap, the spine is made up of three main curves which are cervical, thoracic, and lumbar. Each curve is made up of a multi-segment vertebra that is only divided with a thin layer of an intervertebral disk. Designing a single segment of this multi-joint spinal orthosis that can replicate exactly the dimension of a single vertebra is not an easy task. The dimension of each vertebra is different with increasing height going down from the cervical to the lumbar spine. For the design, modifications and improvements are considered following the new findings from the literature of the spinal biomechanics and reports of user compliances of the commercial existing spinal orthosis. The control system of the entire segment is conducted in the next phase. The 3D model of the final concept design for the lumbar spine is shown in Figure 5.5. Each segment can move up or down within an angle of 25° . This orthosis system will be placed on top of the spine, layered with a smooth surface together with the commercially available back harness to protect the skin from irritation during operation in daily activities.

The proposed system aims to control the spinal curvature of specified spinal segments while allowing other regions to move freely during the daily motion of a person. Different motions require particular spinal curves and segments to be active or passive. Findings and discussions on this subtopic (active and passive spinal segments) are discussed in Chapter 2 (based on literature findings) and Chapter 4 (author's

results) in detail. During operation, the locking rod will either go in (spinal extension) or out (spinal flexion) through the guide hole in each base segment according to the motion of the user as illustrated in Figure 5.6. As the control system triggered the locking function, the locking mechanism will be pulled by a wire and rotated clockwise until it grips the locking rod at its teeth as shown in Figures 5.7 (a) and (b). Thus, it stops the motion of the targeted spinal region that can lead to back pain due to bad posture implementation or excessive load carried. The spin-lock mechanism is designed with a miniature spring that tends to bring it back to the original unlock position if no tension is applied to the wire.

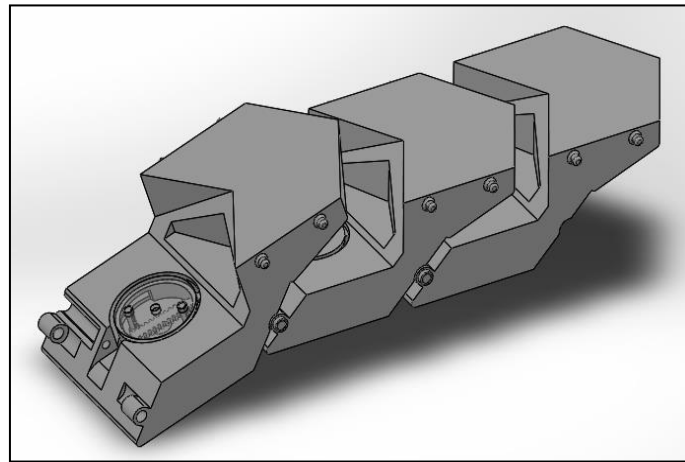


Figure 5. 5 3D models of the final concept design

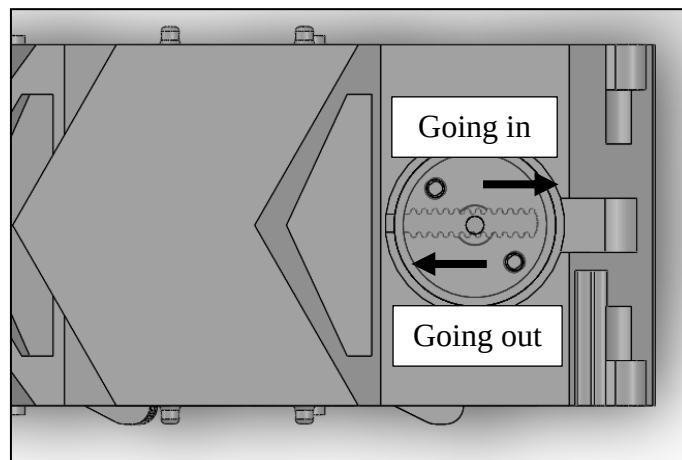


Figure 5. 6 Top views of the model

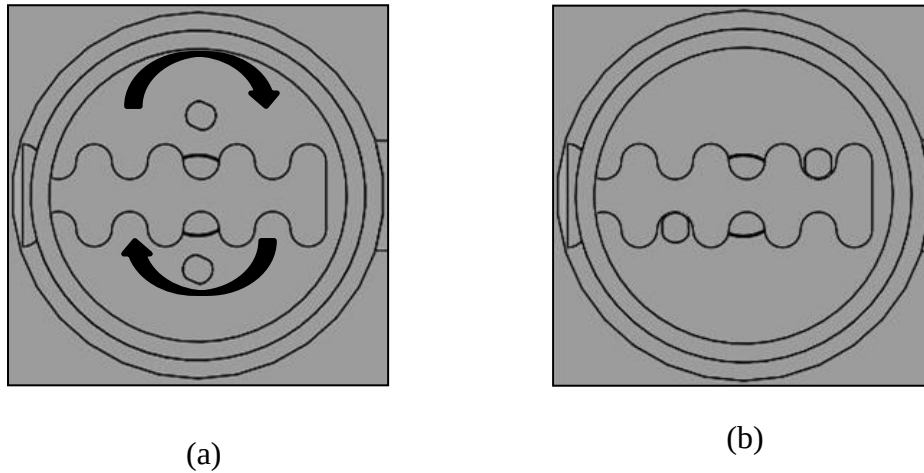


Figure 5. 7 Spin-lock mechanism (a) Unlock and (b) Lock position

After considering the complexity in geometry and difficulties of molding and machining processes that may be included in the fabrication of this concept design, the simplified design for the multiple-joint spinal orthosis is proposed. The simplified design is simpler and fewer parts are used compared to the previous design, without disregarding the allowable range of motion for the two connected segments and the chosen concepts from the concept screening and concept scoring stages. The full assembly of the simplified final concept design proposed is shown in Figure 5.8 below:

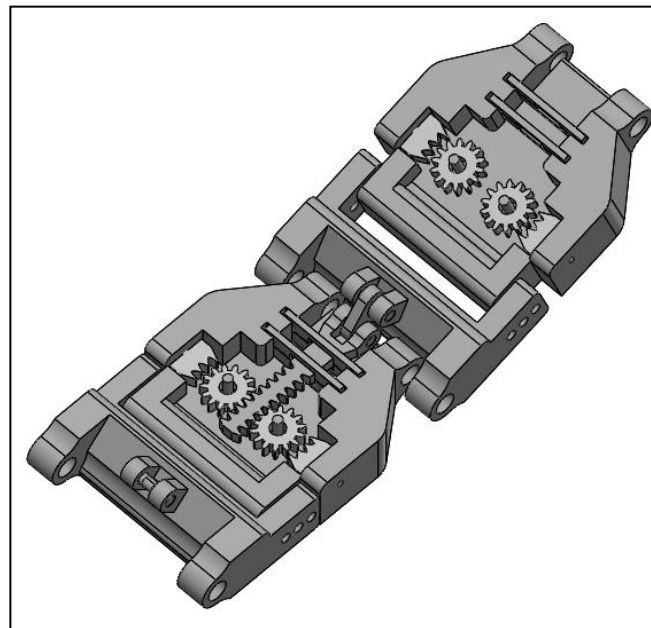


Figure 5. 8 Simplified final concept design

The mechanical system for the simplified design of the multiple-joint spinal orthosis consists of 7 components, which are the front and back base, rack, gears, gear gripper, rods, and flat connector. The main components of the system are shown in Figures 5.9, 5.10, and 5.11:

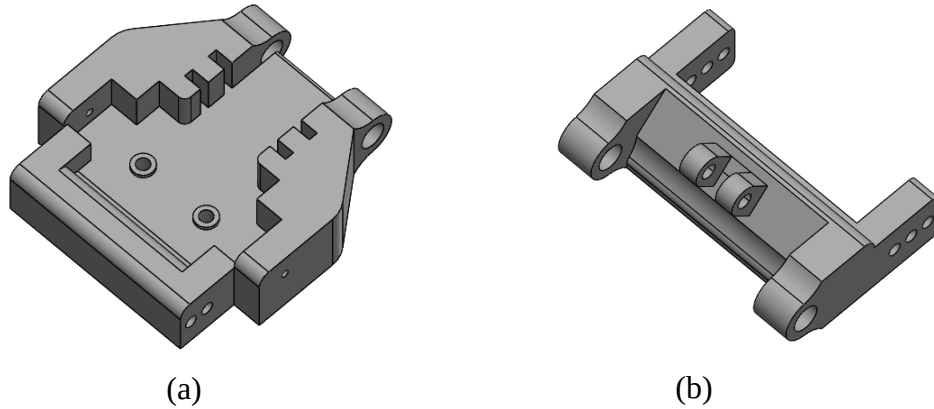


Figure 5. 9 (a) Front base & (b) Back base

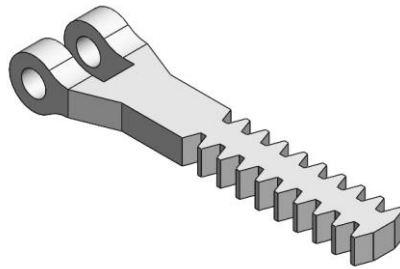


Figure 5. 10 Rack



Figure 5. 11 (a) Gear gripper and (b) Flat-connector

To ensure that the proposed multiple-joint spinal orthosis can sustain the required load in operation, a finite element analysis was performed on the simplified final concept design. The initial load considered in the FEA takes account of the average Malaysian weight recorded which is 58.3 ± 0.1 kg [242] and for the analysis,

the value is rounded to 60 kg. The spine only supports approximately half of the human total weight, thus only 30 kg is used for the analysis and added with another 20 kg of load, considering the user is doing lifting works while wearing the orthosis. The details on the result of the FEA are in Appendix K as it served as the preliminary works. The study came upon an article that looked at the impact of spinal stresses on the fifth lumbar intervertebral discs after reviewing the literature. During weightlifting from the ground, the median greatest resultant force reported was 1375N. Similar to the previous analysis, only half of the load was included, which was about 700N. Due to the vast number of research in the literature for rehabilitation devices utilize commercial software ABAQUS for FEA, such as Ankle-Foot Prosthetic[243], Brace[244-246], biological exoskeleton[247], Constant-Torque Mechanism[248], and Wrist-Hand Orthosis[249, 250] this study used it as well. This choice is also supported by ABAQUS' capability to provide high-accuracy analyses[249]. Subsection 5.3.1 goes through this in-depth as well.

5.2 Complications with the simplified final concept design

Based on the simulation analysis of load applied on the simplified final concept design for the multi-joint spinal orthosis, it can successfully support and sustain the daily basis load acting, especially on the lumbar vertebra. Unfortunately, the design cannot be used for the fabrication process due to many issues such as:

(a) Due to the high complexity of the design proposed, it will eventually increase the number of waste products, labor costs, production time, and the price of components produced, specifically for the gear gripper and rack. Furthermore, the compact and complex geometry required for the gear gripper's design will necessitate considerable caution during fabrication, thus increasing the production cost. Because this component is intended to sustain a significant amount of load during locking, it is not tolerant of failure. Discussions with the technical team and engineer at the local fabrication facility yielded the concerns.

(b) The diameter of the screw holes proposed is too small. Moreover, the size of the screw and pin for the holes are also not commercially available as there is no more production with that dimension based on the survey with local suppliers.

(c) The Front and Back Base designs are overly bulky, requiring a considerable amount of raw materials. Also, if a fault or damage occurred in these components during manufacture or while they were in operation, it might pose problems since reproducing them would take a long time and cost a lot of money. Thus, a detachable feature is preferable to a single bulky design.

A new and final design idea for the manufacturing process has been suggested after evaluating the difficulties and consulting with mechanical specialists, experienced engineers, and suppliers which are discussed in detail in the next section.

5.3 Prototype

In the form of a prototype, the new and final design of the proposed multi-joint spinal orthosis is presented. Figure 5.12 depicts the assembly of the lumbar section prototype. This prototype is made up of 16 components for each segment not including the small components such as screws, nuts, and pins. The number of components is increased to reduce the complexity of the bulky segment of the previous design. The details for each component dimension are in Appendix K The main components used to make each segment as shown in Figures 5.13 to 5.19:

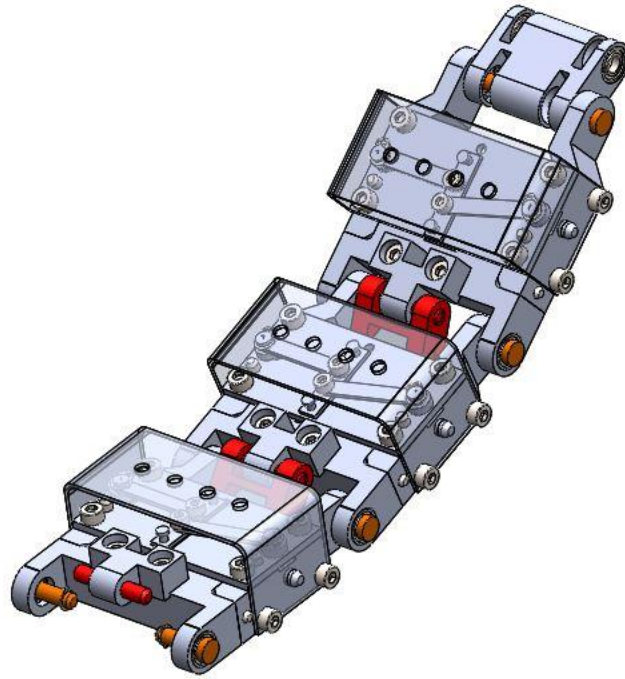


Figure 5. 12 Prototype of the proposed spinal orthosis.

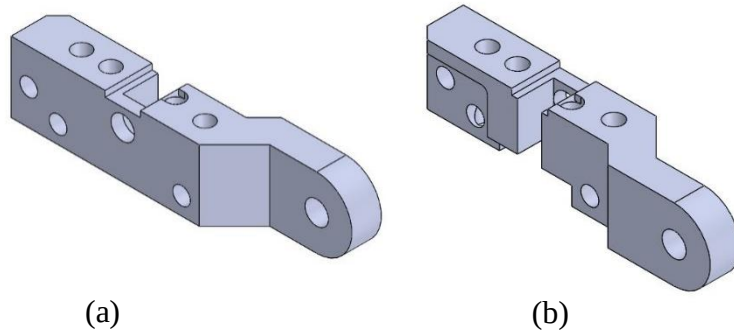


Figure 5. 13 (a) Right and (b) Left base

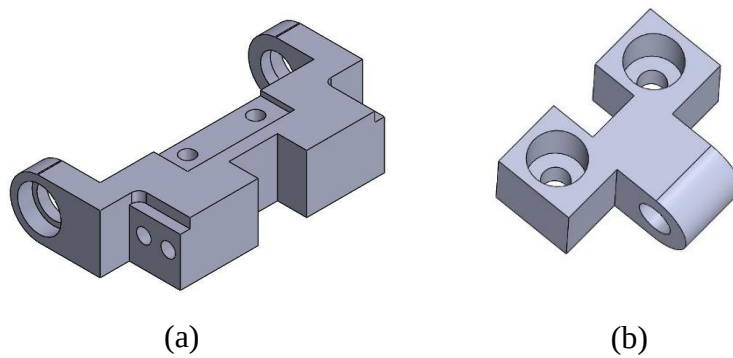


Figure 5. 14 Back base, (a) Main & (b) Support

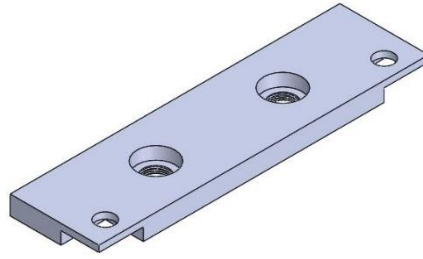
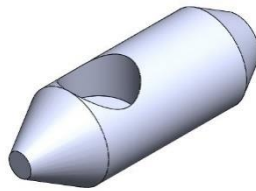
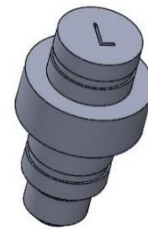


Figure 5. 15 Lower base

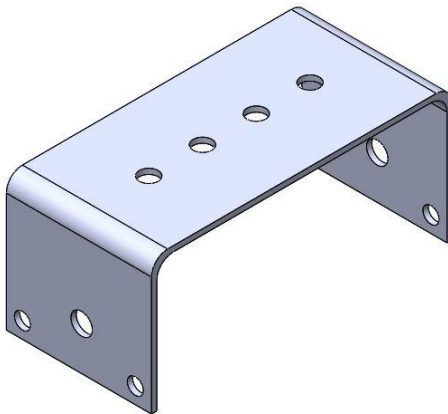


(a)

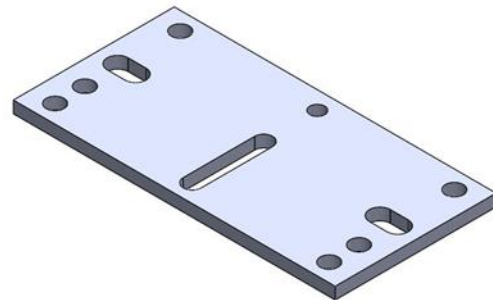


(b)

Figure 5. 16 Lock pin, (a) Pin & (b) Holder

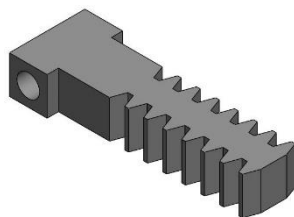


(a)



(b)

Figure 5. 17 Plate, (a) Outer & (b) Inner



(a)



(b)

Figure 5. 18 Rack and pinion, (a) Rack & (b) Pinion

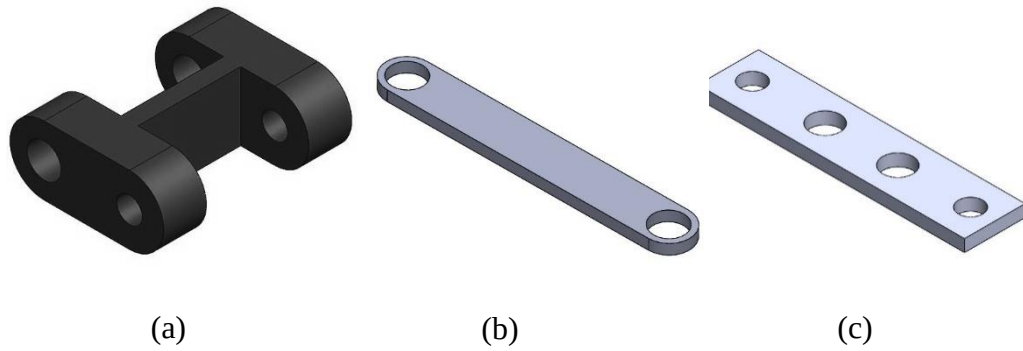


Figure 5. 19 (a) Connector, (b) Linkage & (c) Slider

5.3.1 Finite Element Analysis, FEA

Stronger stress endurance material is used for the components that are designed for locking purposes such as lock pin, rack, and pinion (spur gears). These components are the most susceptible to failure in case of excessive load being applied to it. Considering the high spinal loads acting on fifth lumbar intervertebral discs [3, 65], the median of maximum resultant force recorded was 1375N during weightlifting from the ground. The proposed multiple-joint spinal orthosis was focused on assisting human motion and does not entirely support the human daily motions. Thus, for the simulation FEA analysis using Abaqus, only 50% of the load was taken into consideration, which was approximately 700N. This value was higher compared with the initial 500N applied for the first two spinal orthoses proposed. To ensure the proposed material of these components does not fail during working operation, the maximum stress recorded must fall below its yield strength. The yield strength is defined as the limit at which the predetermined amount of permanent deformation occurs. Components that being studied using FEA analysis were rack, pinion (spur gear), connector, pin, and left base. The boundary condition of each studied component was defined as in Figure 5.20, 5.27, 5.29(a), 5.32(a), and 5.34. 700N load was applied on these components, which then assumed as a uniformly distributed load for the analysis (Figure 5.21, 5.28, 5.29(b), 5.32(b), and 5.35). The results of the regional stress of global seed are indicated as red for the highest stress and dark blue for the lowest for each component studied (Figure 5.22, 5.28, 5.30, 5.33, and 5.36). These results are tabulated in Table 5.1, 5.2, 5.3, 5.4, and 5.5. All components successfully being

analyzed using the material proposed with only the pin lock requires the use of material with higher-yielding stress (change from Stainless steel to Alloy Steel) due to the need for stronger stress-endurance material.

5.3.2 Rack

The initial remeshing approach for the rack displays a localized contour of maximum stress within the specified size of global seed used which is highlighted in the green box in Figure 5.22. The drawback of applying this approach was that as the size of the global seed increases, the number of elements and processing time will increase dramatically. Thus, to eliminate such waste of time, biased meshing was used. This approach was used especially for the critical region to minimize the number of elements thus enhancing the result's accuracy as displayed in Figures 5.23 and 5.24. Different sizes of global seed applied to the separated region and max stress recorded are shown in Table 5.2 and 5.3 below. By using the results obtained from the global seed, a smaller regional seed was applied to the region with bright contour (red indicates maximum stress) which was also the critical region. The results for the rack are shown in Figures 5.25 and 5.26. Second and third-degree biased remeshing was only simulated for the rack, connector, and base due to the complex geometry of these components compared with the other components and they are more prone to failure.

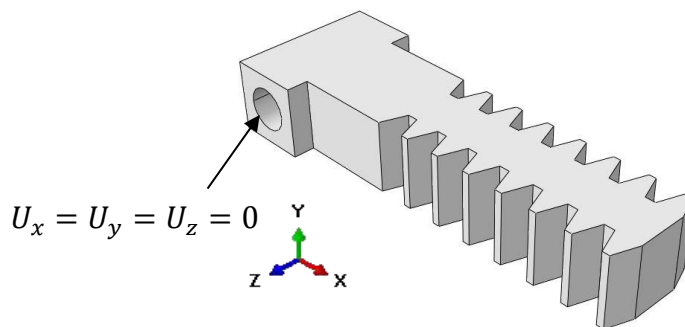


Figure 5. 20 Boundary condition for Rack.

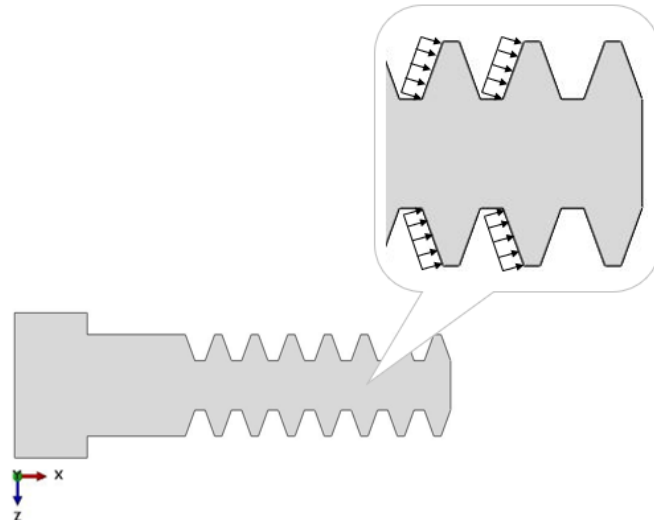


Figure 5. 21 Applied load on Rack.

Table 5. 1 FEA Meshing results for rack using global seed

Meshing Stage	No. of elements	Max Stress (MPa)	Global Seed (mm)	Relative CPU time (s)
1	8155	118.00	1.00	3.20
2	18213	127.10	0.75	10.60
3	44806	140.70	0.50	40.40
4	93326	215.57	0.40	169.70
5	177432	228.03	0.30	500.90

Table 5. 2 FEA Biased meshing results for the rack with 2 different sizes

Meshing Stage	No. of elements	Max Stress (MPa)	Global Seed (mm)		Relative CPU time (s)
			Large	Small	
1	7681	140.80	1.50	0.50	2.80
2	19393	204.60	1.50	0.40	13.30
3	26325	207.30	1.50	0.30	23.40
4	95151	348.90	1.50	0.20	238.20

Table 5. 3 FEA Biased meshing results for the rack with 3 different sizes

Meshing Stage	No. of elements	Max Stress (MPa)	Global Seed (mm)			Relative CPU time (s)
			Large	Middle	Small	
1	24340	270.44	1.50	0.80	0.25	26.80
2	43084	343.40	1.50	0.80	0.20	83.10
3	59983	398.00	1.50	0.80	0.15	104.10
4	102407	328.60	1.50	0.50	0.10	351.20

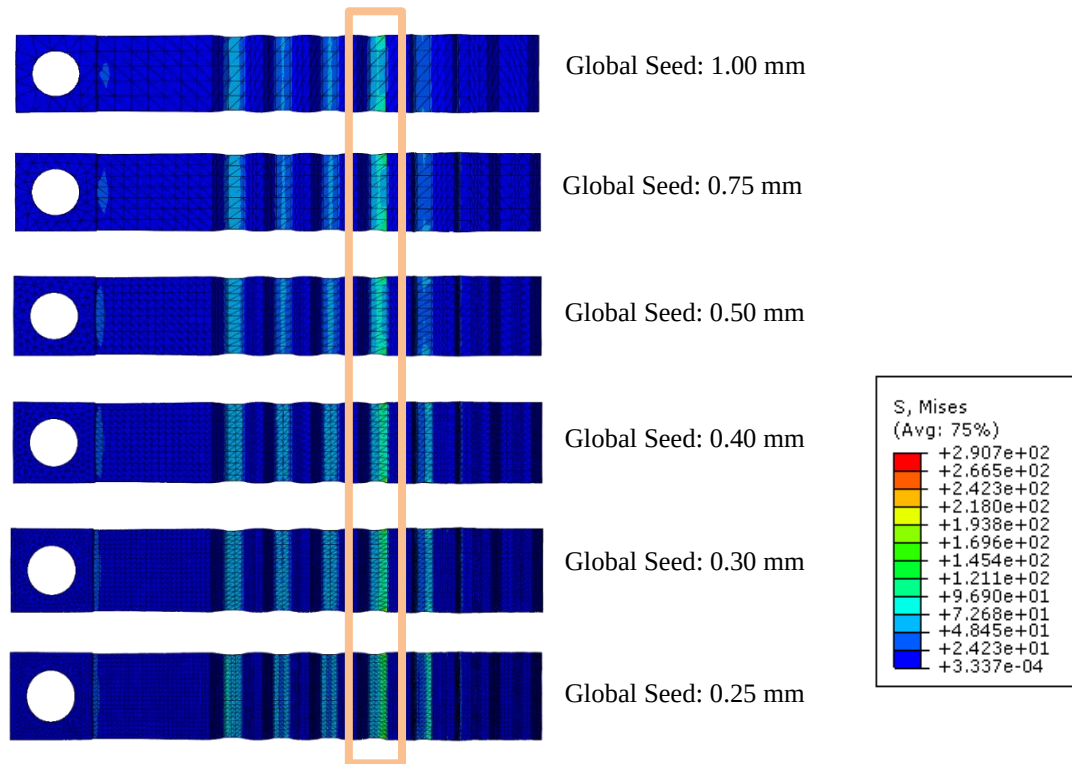


Figure 5. 22 Different sizes of global seed (in mm) applied on rack

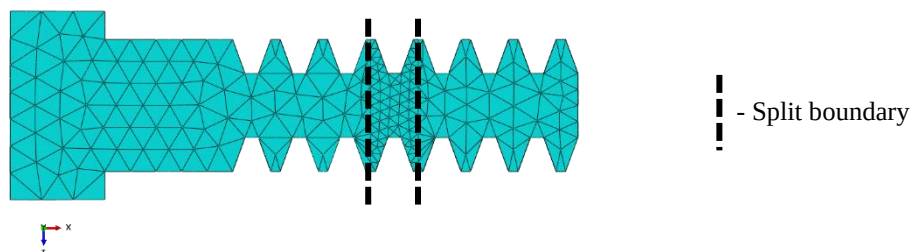


Figure 5. 23 FEA Biased meshing applied on the rack (2 different sizes)
Large and small seed: 1.50 and 0.50 mm, respectively

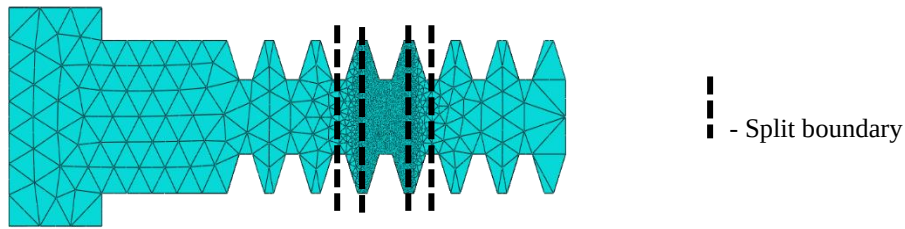


Figure 5. 24 FEA Biased meshing applied on the rack (3 different sizes)
Large, middle, and small seed: 1.50, 0.50, and 0.10 mm, respectively.

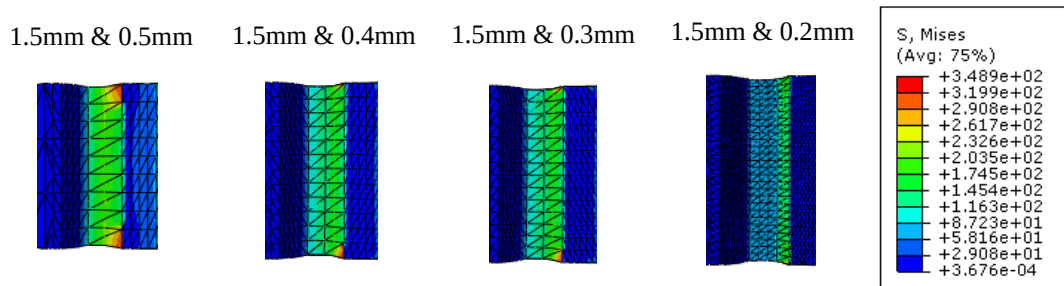


Figure 5. 25 Meshing result for 2 different sizes

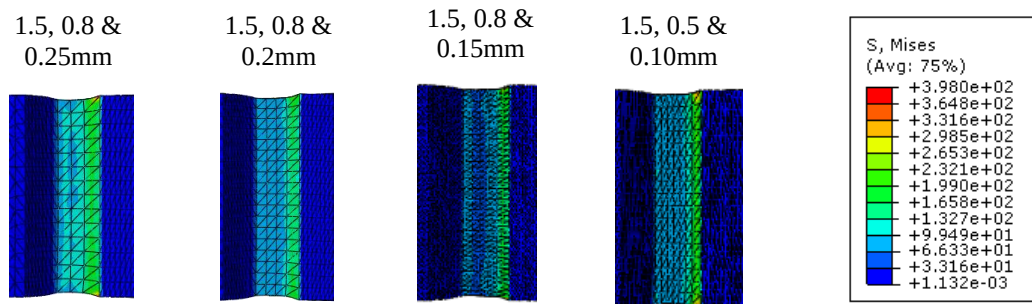


Figure 5. 26 Meshing result for 3 different sizes

5.3.3 Pinion (spur gear)

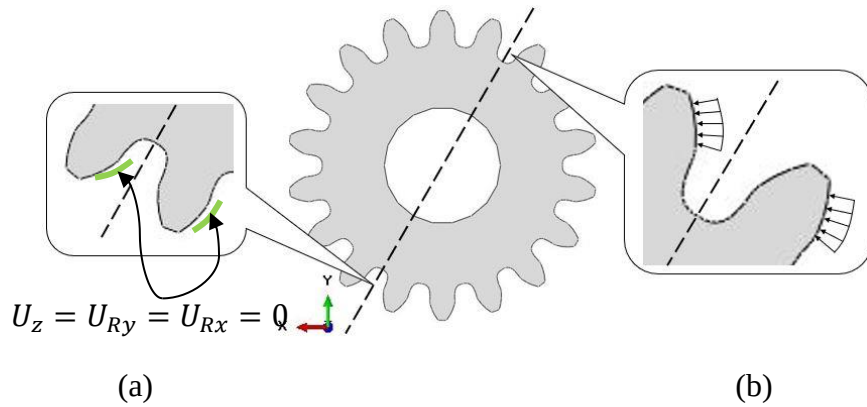


Figure 5.27 Pinion (spur gear). a) Boundary condition and (b) Applied load

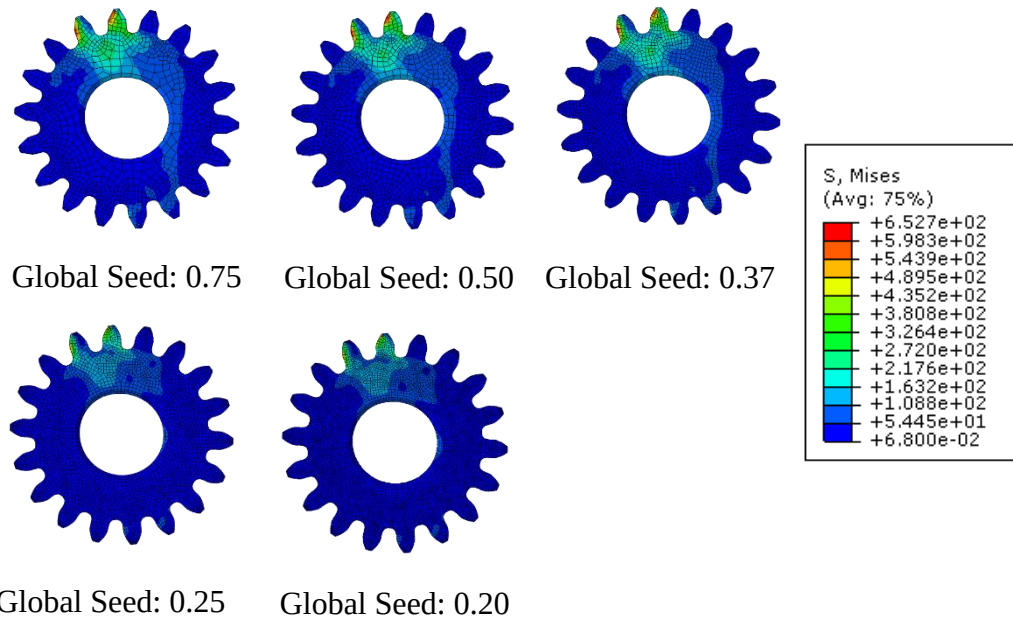


Figure 5.28 Different sizes of global seed (in mm) applied on the pinion gear

Table 5.4 FEA Meshing results for pinion gear using global seed

Meshing Stage	No. of elements	Max Stress (MPa)	Global Seed (mm)	Relative CPU time (s)
1	8351	398.40	0.75	1.60
2	14750	438.50	0.50	4.70
3	30730	445.90	0.37	8.60
4	60580	650.40	0.25	36.00
5	134675	652.70	0.20	136.60

5.3.4 Connector

Similar to the rack, there was a need to analyze the connector with a biased remeshing approach as the peak stresses are in specific regions. Applying the global seed in the first place has successfully pinpointed these regions, thus the biased remeshing can increase the analysis accuracy by focusing only on the area of interest as in Figure 5.31. Applying the biased remeshing of 0.30 and 1.20 mm resulted in 129205 elements and maximum stress of 122.30 MPa as in Table 5.6.

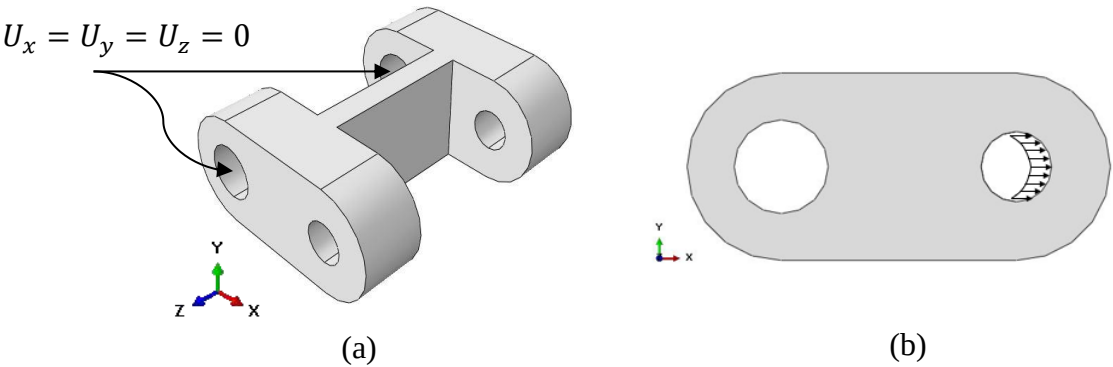


Figure 5. 29 Connector. a) Boundary condition and (b) Applied load

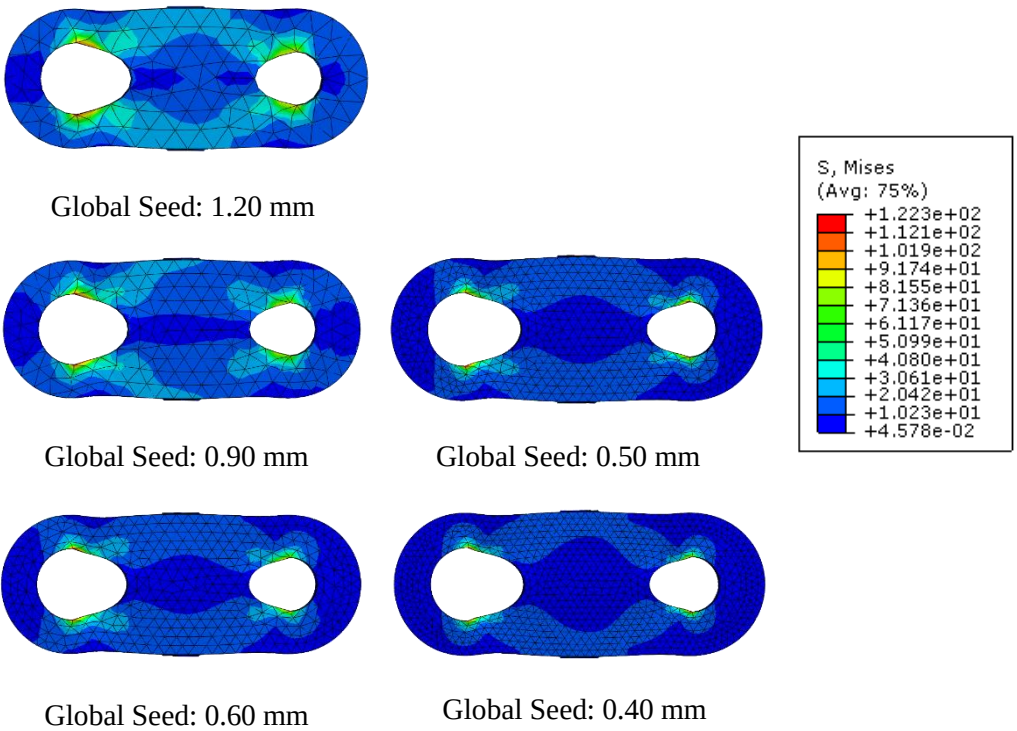


Figure 5. 30 Different sizes of global seed (in mm) applied on connector

Table 5. 5 FEA Meshing results for connector using global seed

Meshing Stage	No. of elements	Max Stress (MPa)	Global Seed (mm)	Relative CPU time (s)
1	5654	60.11	1.20	1.70
2	14064	71.53	0.90	6.30
3	43548	95.41	0.60	39.30
4	71582	101.50	0.50	63.20
5	139574	122.30	0.40	310.50

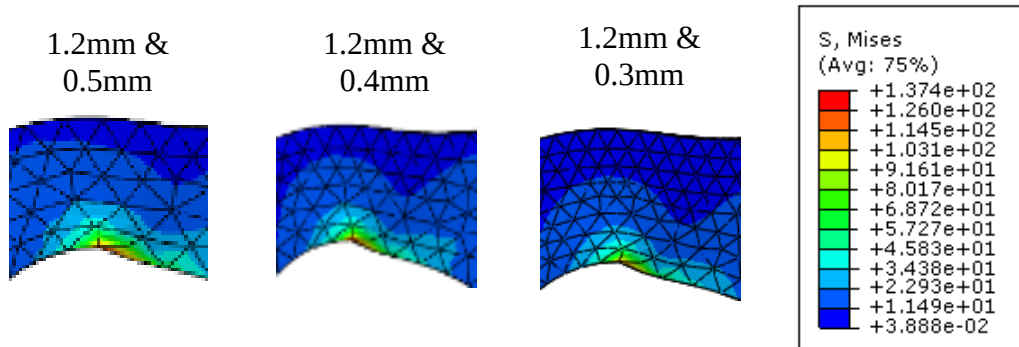


Figure 5. 31 Meshing result for 2 sizes (connector)

Table 5. 6 FEA Biased meshing results for the connector with 2 different sizes

Meshing Stage	No. of elements	Max Stress (MPa)	Global Seed (mm)		Relative CPU time (s)
			Large	Small	
1	38308	106.00	1.20	0.50	31.10
2	62698	119.90	1.20	0.40	53.00
3	129205	137.40	1.20	0.30	170.10

5.3.5 Locking pin

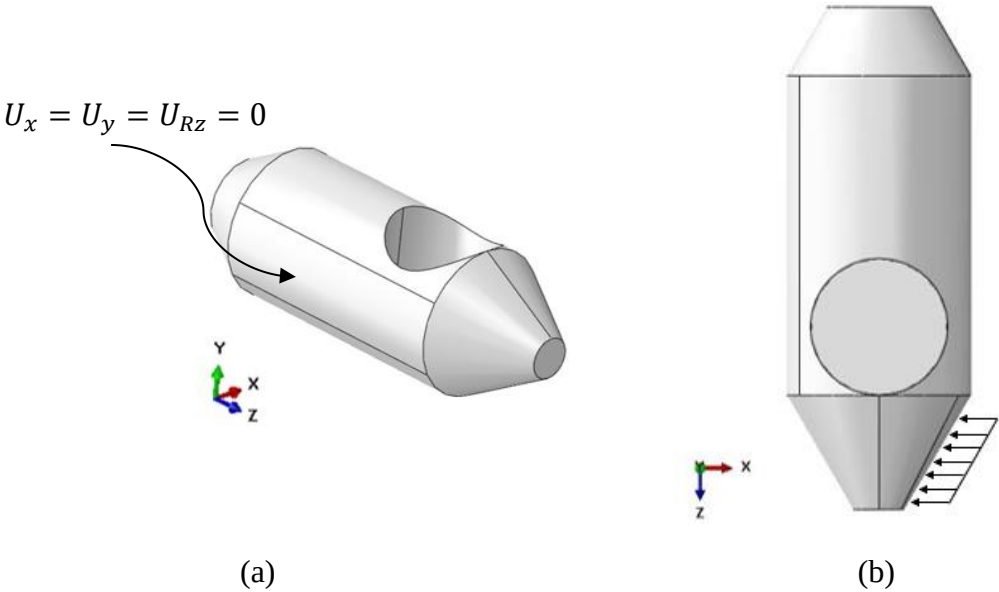


Figure 5.32 Lock pin. a) Boundary condition and (b) Applied load

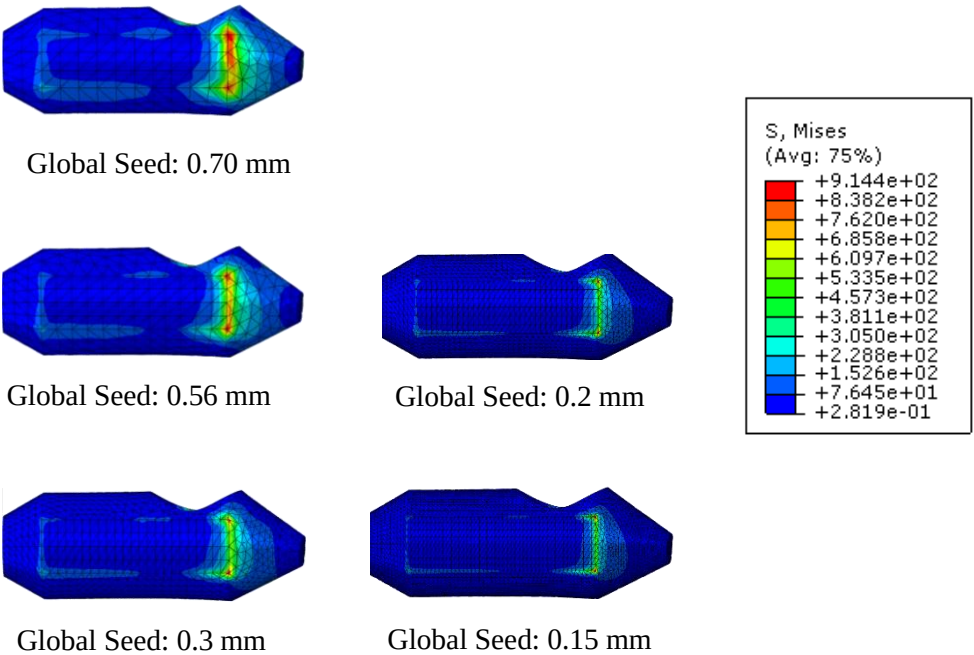


Figure 5.33 Different sizes of global seed (in mm) applied on locking pin

Table 5. 7 FEA Meshing results for locking pin using global seed

Meshing Stage	No. of elements	Max Stress (MPa)	Global Seed (mm)	Relative CPU time (s)
1	2183	319.40	0.70	0.80
2	4377	410.60	0.56	2.20
3	23020	571.20	0.30	24.40
4	67682	796.80	0.20	118.00
5	132455	914.40	0.15	526.90

5.3.6 Left base

From the global seed meshing analysis, the most critical region that showed peak stress on the base has successfully been identified as in Figure 5.37. Using biased remeshing on this region with 0.35 and 1.60 mm global seed (Figure 5.38), the maximum stress recorded was 90.51 MPa as in Table 5.9.

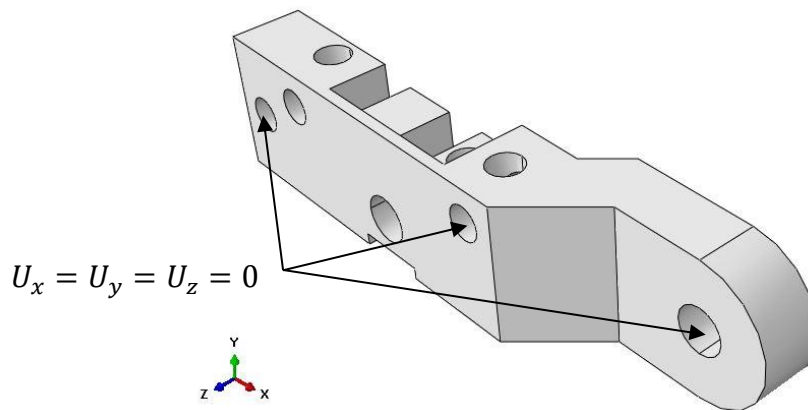


Figure 5. 34 Boundary condition for Left base.

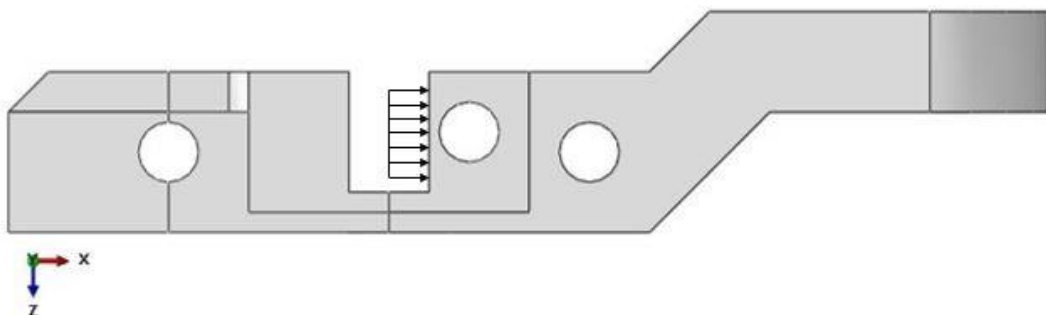


Figure 5. 35 Applied loads for Left base.

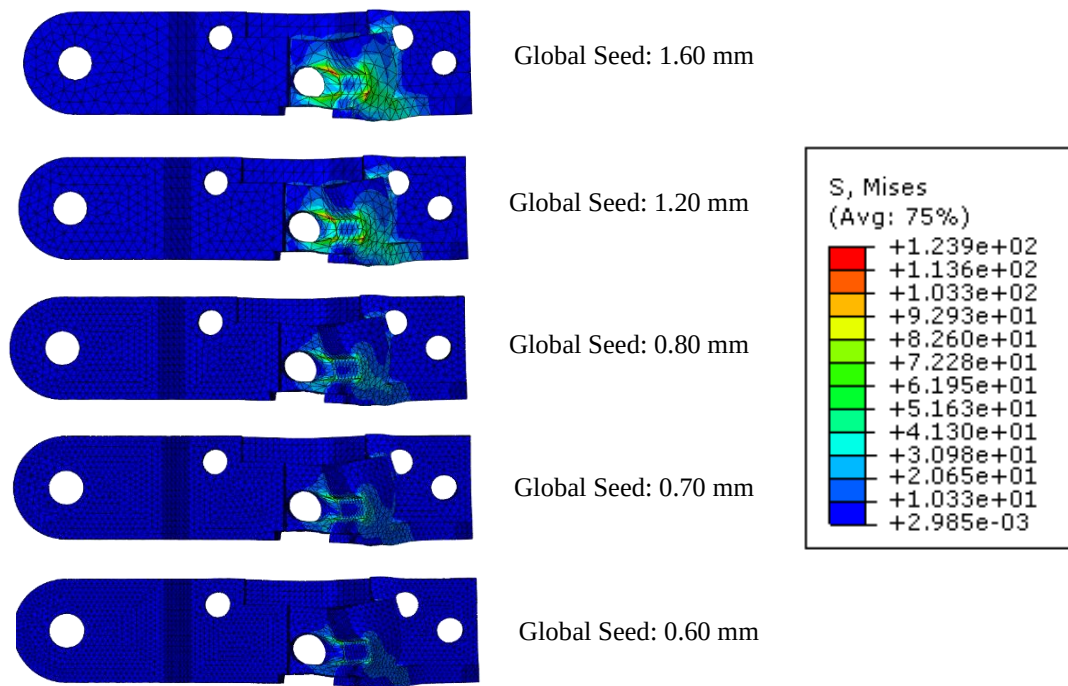


Figure 5. 36 Different sizes of global seed (in mm) applied on left base

Table 5. 8 FEA Meshing results for the left base using global seed

Meshing Stage	No. of elements	Max Stress (MPa)	Global Seed (mm)	Relative CPU time (s)
1	9289	36.09	1.60	4.00
2	16180	43.78	1.20	9.30
3	50676	81.08	0.80	34.70
4	69261	93.01	0.70	74.80
5	104780	123.90	0.60	182.80

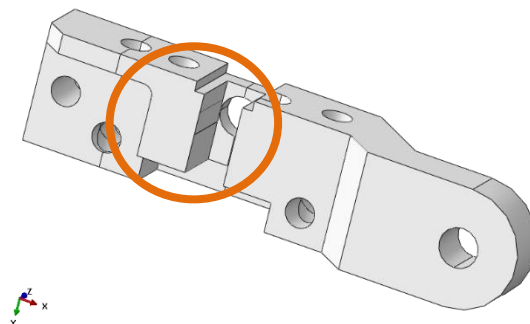


Figure 5. 37 Encircled critical stress contour

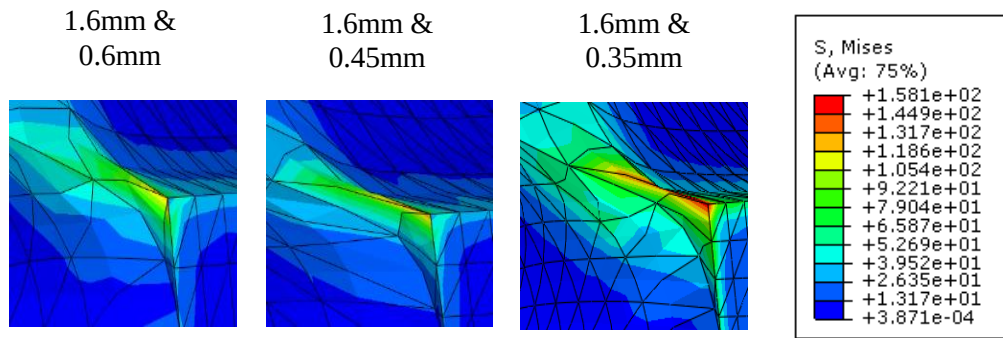


Figure 5. 38 FEA biased mesh for Left Base

Table 5. 9 FEA Biased meshing results for the left base with 2 different sizes

Meshing Stage	No. of elements	Max Stress (MPa)	Global Seed (mm)		Relative CPU time (s)
			Large	Small	
1	31946	117.70	1.60	0.60	29.00
2	61519	158.10	1.60	0.45	97.90
3	110981	90.51	1.60	0.35	340.70

5.4 Summary of the FEA and chosen material

All components successfully being analyzed using the material proposed with only the pin lock requires the use of material with higher-yielding stress (change from Stainless steel to Alloy Steel) due to the need for stronger stress-endurance material (Table 5.10). Mesh convergence in a finite element analysis defines how many elements are necessary for a model to ensure that the findings of the analysis are unaffected by mesh size changes. However, the analysis in this study did not show any mesh convergence characteristics for all components analyzed. The complex geometry of the components may be responsible for this result. For the solution, this study implements the approach used by Afripin [251], which stated that the convergence criterion is based on values of equivalent plastic strain (PEEQ) in the control volume, predicted for the last two successive runs. Similarly, if the localized stress contour appeared in the last two runs, the convergence criterion is deemed to have been met. For the components analyzed in ABAQUS, all of them have successfully achieved localized stress contour, even after conducting the biased meshing.

Table 5. 10 Summary of the material chosen for the components

Component	Material	Max Stress, MPa	Max Yield strength, MPa
Rack	Alloy Steel	398.00	2400 MPa
Pinion (spur gear)	Brass	652.70	683 MPa
Connector	Stainless Steel	137.40	215 MPa
Locking Pin	Alloy Steel	914.40	2400 MPa
Left, Right, and Back Base	Aluminium 6061-T6	158.10	276 MPa

5.5 Control Implementation

This section is divided into two main sections, which are on the prototype and control system for the proposed multiple-joint spinal orthosis. The first subsection discussed in the depth the mechanism and modifications that are necessary to ensure the workability of the mechanical components of the orthosis. The next subsection talks about the implementation of a control system based on the author's findings on the active-passive spinal segments in completing critical daily activities as discussed in chapter 4.

5.5.1 Modifications to the prototype's components

As discussed in detail in subsection 5.3, the prototype of the orthosis was used for the fabrication process. However, due to the limited research budget, only three lumbar segments able to be fabricated. Furthermore, the prototype does not operate as smoothly as it should. Several components' clearance gaps for slider and rotating mechanisms were not established prior to manufacture, resulting in mechanical friction. Several modifications of selected components have been done to ensure workability and improve the efficiency of the mechanism. Also, the locking system has been improved using a simplified mechanical mechanism.

5.5.1.1 Outer Plate

The main purpose of the outer plate was to protect the locking mechanism from external pressure or load. This component was also used to house the string that was

used to activate or deactivate the locking mechanism. A 4mm diameter hole was made at each side of the outer plate to allow a clearance gap for locking pin during unlocking position as in Figure 6.1. However, during the assembly process, friction between the lock pin and the outer cover was present, thus reducing the efficiency of the locking system. To fix this, the diameter was then increased to 8mm. Two M3 holes were made at both sides of the outer plate, specifically at lower corners which are used to attach the outer plate to each segment. However, the presence of too many M3 threads has made the assembly process complicated. Furthermore, because the plate is only 1mm thick, the screw thread is not critical. Thus, 4mm diameter holes were drilled to eliminate all the M3 thread thus making the assembly process quicker. The improved version of the outer plate is shown in Figure 6.2.

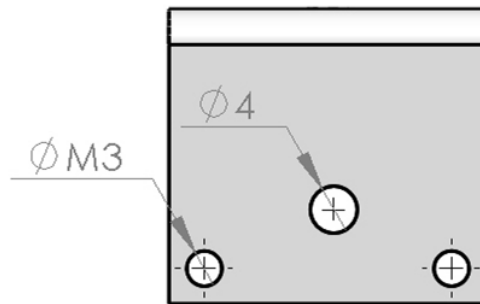


Figure 6. 1 Initial dimension used for the outer plate

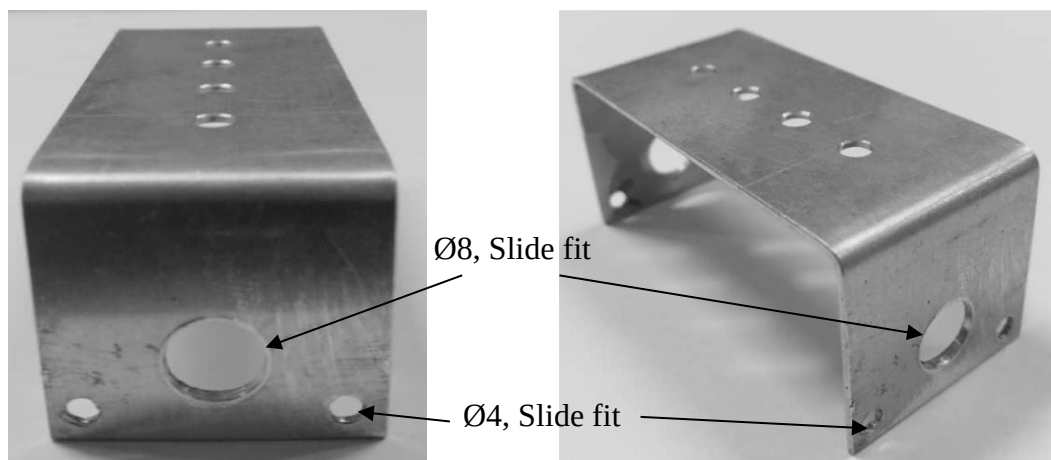


Figure 6. 2 Improved version of outer cover

5.5.1.2 Inner plate

This component is responsible for holding the Left, Right, and Back Base in place as well as providing the locking mechanism's slider path. The issue that arises for the inner plate was the same as the outer plate, which was the problem with the thread of the M3 hole. Initially, four M3 holes (Figure 6.3(a)) were drilled at each corner of the plate to fix it with the left and right base using Hex cap M3 screws. These threads have then been removed by drilling them over with 4mm diameter holes as shown in Figure 6.3(b). To bind together with Left, Right, and Back Bases, the M3 thread positioned at the Left and Right Bases were used.

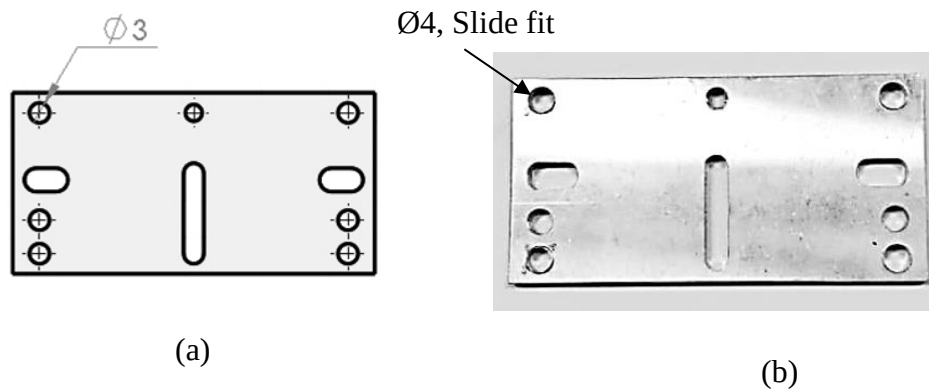


Figure 6. 3 Inner plate (a) Initial and (b) Improved version

5.5.1.3 Spur gear rod

Initially, the installation of the spur gear rod to the base body was designed with a tight-fit feature. However, the rod keeps on uprooted from its hole due to the loose hole's diameter. This component is essential for keeping the spur gears in position and ensuring that the locking mechanism operates as smoothly as possible. Additionally, the outer and inner diameter of the rod and the gear respectively were fabricated with a dimension of 6mm without considering the friction gap. During operation, the spur gear should be able to rotate freely. The absence of the friction gap disturbs the efficiency of the rotation due to the presence of great friction with the surface. To solve the issue, a new rod with a smaller diameter, which was 5.8mm was used. This rod was also drilled from top to end and tapped for an M3 screw. An M3 screw with 12mm length was used to fix both rods to each base body as shown in

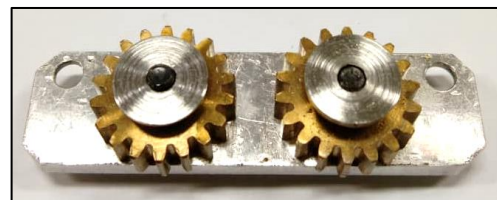
Figure 6.4 below. The rods will be firmly attached to the Base as a result of this. The assembly of the rods, spur gears base was shown in Figures 6.5 (a) and (b).



Figure 6. 4 Modified rod and 12mm length M3 screw used.



(a)



(b)

Figure 6. 5 Assembly of the modified rods, spur gears, and base.
(a)Side and (b) Top view

5.5.2 Lock-Unlock mechanism

The Lock-Unlock mechanism of the fabricated prototype, unfortunately, failed to operate efficiently. One of the reasons was due to the manufacturing defects for the Left and Right bases. The hole for the lock pin was misaligned as shown in Figure 6.6 below. The gap distance was designed for lock pin moves in and out was also miscalculated by the factory. The initial distance based on the schematic diagram in the Appendix was 4mm in length as in Figure 6.7. However, the inner surface of the gap was cut unevenly, creating a rough sliding surface for the pin. The issues have successfully been solved by cutting a new hole in the correct position and increase the gap distance by 4.5mm.



Figure 6. 6 Misaligned hole for the pin (Correct hole's position is circled in black dash)

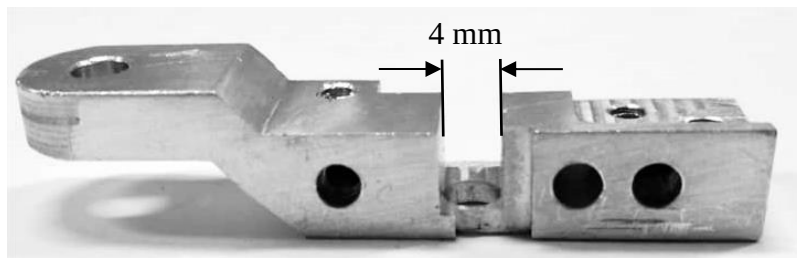


Figure 6. 7 Gap distance for the lock pin

Another issue that arises was the problem with the linkage length. Initially, two linkages with different dimensions were proposed for each segment as shown in Figure 6.8 below. The lengths proposed for linkage A and B were 23 mm and 24 mm respectively. The difference in length was initially to prevent the tilting of the pin holder if the linkages are stacked together, which can disrupt the smoothness of the motion of the lock pins. The cross-sectional front view of the components of the locking mechanism is shown in Figure 6.9 below. However, only one side of each segment's locking mechanism works perfectly during the testing process of the prototype, specifically when using the shorter linkage. Thus, the design was revised and a 23 mm length was set for both sides of the linkages. An extension spring was also installed to ensure all lock pins are always in the release position or Unlock mode. If the Lock mode is activated, the screw that connects both linkages will be pulled downward by a string connected to the Torque servo, thus extending the spring up to 5mm in length. The linkages will be back to their initial position if no torque is applied with the help of the extension spring. The final working locking mechanism is shown in Figure 6.10 for Lock-Unlock mode.

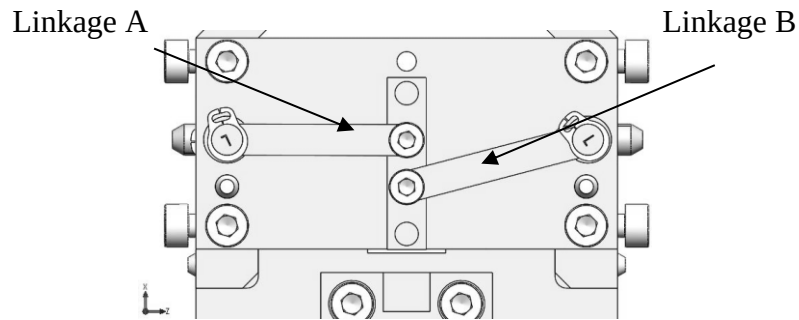


Figure 6. 8 Top view of the linkages used for locking mechanism

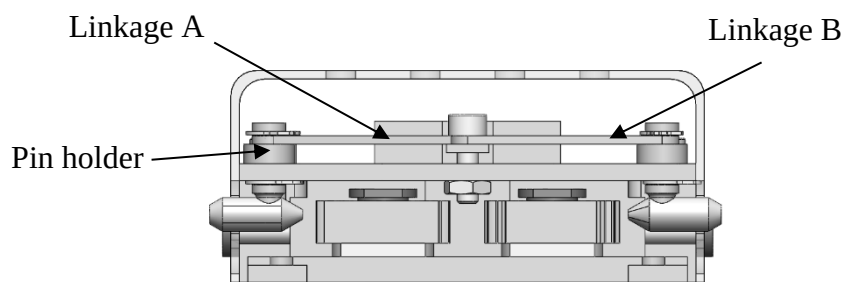


Figure 6. 9 Cross-sectional view of the assembly of the locking mechanism components

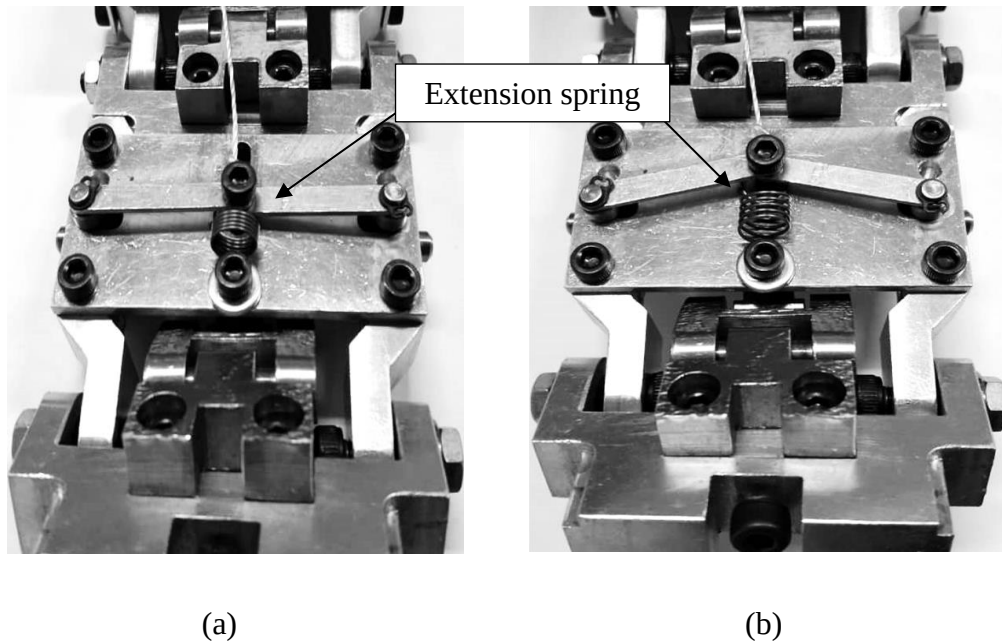


Figure 6. 10 Top view of the locking mechanism. (a) Unlock and (b) Lock mode

5.5.3 Control System

For this study, only three lumbar segments can be manufactured, as originally stated due to time and financial constraints. This research uses a simple control system to run the prototype. The input for this control system is from the changes of the angular displacement of the hip and knee, measured using the modified potentiometers. The reference values of these angles were obtained from the kinematic analysis, as discussed in detail in subsection 4.6. If the real-time angles for the hip and knee are within the Based on these inputs provided to the Arduino, three MG90S servos act as a mode changer that makes the "Unlock or Lock" decision. Servos A, B, and C are set for the Upper (T10-L1), Middle (L1-3), and Lower (L3-5) lumbar segments respectively. A mode for a single segment is determined by each servo. For pulling the string that will lock the required segments, a high-torque servo, MG995 is used. This small yet efficient servo performs well as the weight aspect of a user-friendly orthosis is taken into account. The control system's flowchart is shown in Figure 6.11.

The changes in spinal segmental kinematic of the thoracic and lumbar regions are compared to the reference values of hip and knee angles utilized in this control system. The Lock mode will start if the thoracic or lumbar region exhibited passive kinematic. In the same way, if the segment has active kinematics, the Active mode will be enabled. If the current hip angle measurement is within the range of the Lock mode of the reference angles for the first condition, the control system will send a signal to Servo A, B, C, or all three servos to activate. If the current value is not within the value's range, the control system will command all three servos to stay still. If at least one of the three small servos is triggered, the torque servo will be engaged as well. The torque servo, on the other hand, will not be triggered if there are no small servos activated. The following is a brief overview of the conditions:

Condition 1 (Servo A, B, and C):

- If the current value of the hip angle is within Lock mode (Passive segment), give the output “Yes”.
- If the current value of the hip angle is within Unlock mode (Active segment), send the “No” output.

Condition 2 (Torque Servo):

- If at least one of the servos gives “Yes” output (Lock mode), give the output “Yes”.
- If none of the servos give “Yes” output (mode), give the output “No”.

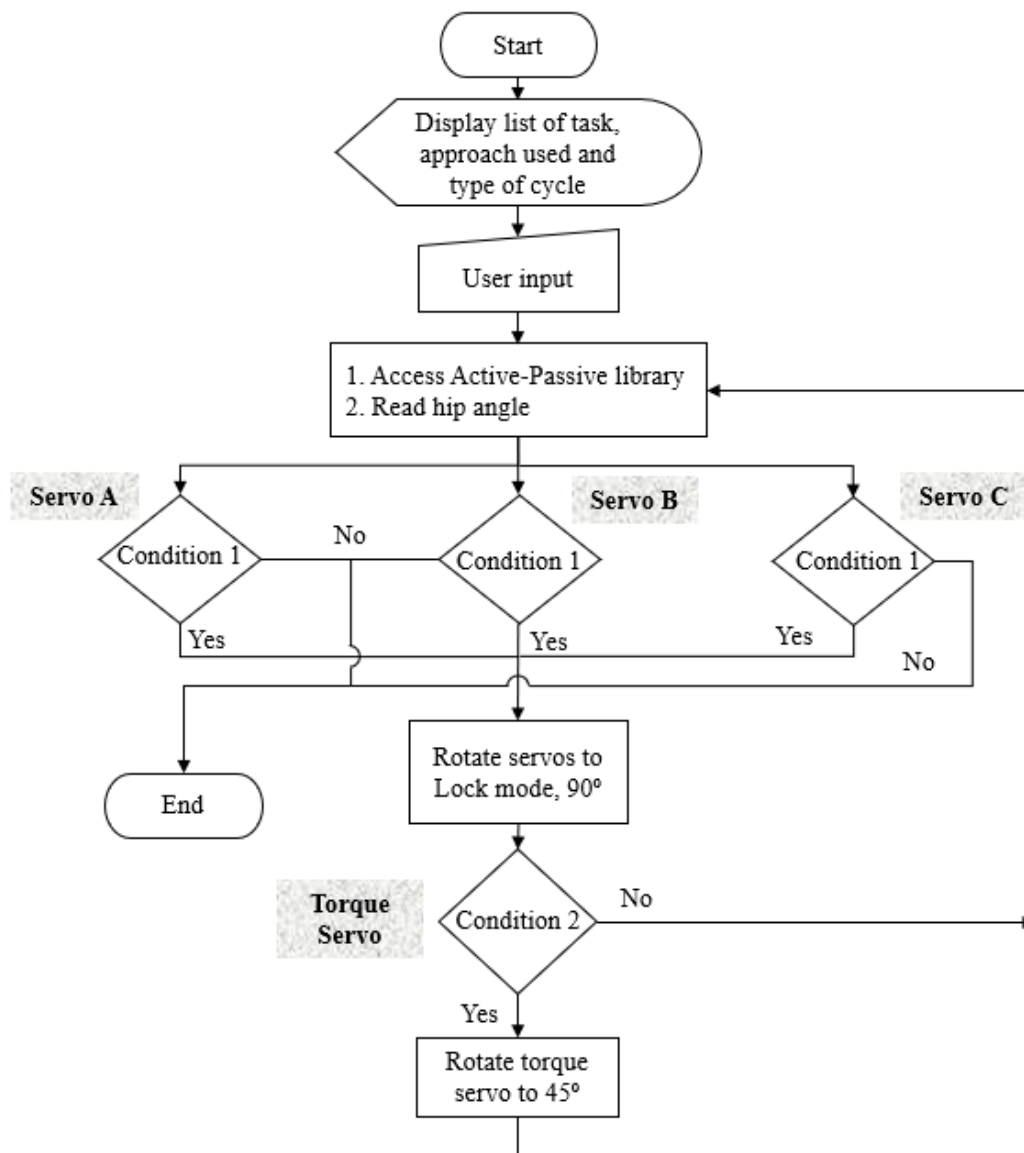


Figure 6. 11 Flowchart of the control system of the proposed multiple-joint spinal orthosis

CHAPTER 6

CONCLUSION AND RECOMMENDATION

6.1 Concluding Remarks

This research has investigated the postural movement of the spinal kinematics in critical daily activities that displayed different contributions. This thesis also presented the findings on the active and passive spinal segments and their contribution to the cervical, thoracic, and lumbar curves which are the key objectives of the study. Using the findings, this research has successfully contributed to a data profile of the active-passive spinal segments which is significantly important in understanding the complex spinal kinematics. This thesis found the contribution of assuming thoracic and lumbar spine as a single and multi-segment is substantially different depending on the different activities studied. Thus, highlighting the importance of implementing the segmental approach in understanding the actual spinal kinematics. Utilizing the findings, a multi-joint spinal orthosis is proposed and fabricated after the FEA analysis is conducted to choose the suitable material. The initial integrated mechatronics system also has been developed to demonstrate the locking mechanism of the proposed multi-joint spinal orthosis which implements the hip and knee angle as the input. The design of the orthosis implemented a simple locking mechanism yet effective in controlling specific spinal segments. Unlike the current multiple joint spinal orthoses which are designed only up to 3 segments, the proposed multi-joint spinal orthosis can control and assist the user's spinal curves which are made up of many segments, covering the lumbar up to cervical regions.

6.2 Contribution to knowledge

This study can contribute to the rehabilitation therapy of the patients that suffer from back pain, muscle training, and human physical support in assisting the manual worker during lifting a heavy load. Moreover, it shows a promising future for robotic rehabilitation applications.

6.3 Future Works

The data profile of the active-passive spinal segments in this research is obtained only within healthy subjects. There is an urgent need to understand the difference in spinal segmental kinematics of both healthy and LBP patients. Thus, future works should focus on the extensive research of the active-passive data profile of people with back pain. As mentioned in Chapter 5, this research can fabricate the prototype only for lumbar segments due to financial and time constraints. Thus, the future study will continue the fabrication phase for the cervical and thoracic multi-segment with the added control for lateral bending and rotation for both left and right directions. Develop predictive modeling for intelligent dynamic orthosis that can predict the motion of the user and give assistance automatically. The clinical study for the completed spinal orthosis that covers from lumbar up to cervical regions will be conducted to examine the efficiency of the proposed multi-joint spinal orthosis.

REFERENCES

1. Wong, E. *Pain in the back*. 2012 12/07/2013 12/10/2015]; Available from: <http://www.thestar.com.my/Lifestyle/Health/2012/08/19/Pain-in-the-back/>.
2. Sizer, P.S., J.M. Brismée, and C. Cook, *Medical screening for red flags in the diagnosis and management of musculoskeletal spine pain*. Pain Practice, 2007. 7(1): p. 53-71.
3. Rohlmann, A., et al., *Activities of everyday life with high spinal loads*. Plos One, 2014. 9(5): p. e98510-e98510.
4. Panjabi, M.M., *Clinical spinal instability and low back pain*. Journal of electromyography and kinesiology, 2003. 13(4): p. 371-379.
5. White, A.A. and M.M. Panjabi, *Clinical Biomechanics of the Spine*. 1990: Lippincott.
6. Kwok, A., et al., *Prevalence of vertebral fracture in Asian men and women: comparison between Hong Kong, Thailand, Indonesia and Japan*. Public Health, 2012. 126(6): p. 523-531.
7. Hoy, D., et al., *A systematic review of the global prevalence of low back pain*. Arthritis & Rheumatism, 2012. 64(6): p. 2028-2037.
8. Damanhuri, Z., et al., *Low back pain among office workers in a public university in Malaysia*. International Journal of Public Health and Clinical Sciences, 2014. 1(1): p. 99-108.
9. Lonik, E.A.T., et al., *Prevalence of low-back pain among public ambulance workers in Kelantan, Malaysia*. Malays J Public Health Med, 2017. 17: p. 126-135.
10. Sivasampu, S., et al., *National Medical Care Statistics (NMCS) 2012*. Kuala Lumpur: National Clinical Research Centre, 2014.
11. Veerapen, K., R.D. Wigley, and H. Valkenburg, *Musculoskeletal pain in Malaysia: a COPCORD survey*. The Journal of Rheumatology, 2007. 34(1): p. 207-13.
12. Aziz, R.A., et al., *Effect of Working Posture on Back Pain Occurrence among Electronic Workers in Malaysia*. Procedia Manufacturing, 2015. 2: p. 296-300.
13. AL-Dubai, S.A.R., et al., *Prevalence and determinants of low back pain among taxi drivers in Malaysia. A cross sectional study*. Journal of Advanced Medical Research, 2012. 2(4): p. 129-43.
14. Hani, S. and S. Liew, *The views and experiences of Malaysian primary care doctors in managing patients with chronic low back pain: a qualitative study*. Malaysian Family Physician: the Official Journal of the Academy of Family Physicians of Malaysia, 2018. 13(1): p. 18.
15. Riihimäki, H., *Low-back pain, its origin and risk indicators*. Scandinavian journal of work, environment & health, 1991: p. 81-90.
16. Craig, B.N., et al., *Occupational risk factors and back injury*. International Journal Of Occupational Safety And Ergonomics: JOSE, 2013. 19(3): p. 335-345.
17. Kuo, Y.-L., E.A. Tully, and M.P. Galea, *Video analysis of sagittal spinal posture in healthy young and older adults*. Journal of manipulative and physiological therapeutics, 2009. 32(3): p. 210-215.

18. Grandjean, E. and W. Hünting, *Ergonomics of posture—Review of various problems of standing and sitting posture*. Applied Ergonomics, 1977. **8**(3): p. 135-140.
19. Sung, P.S., P. Danial, and D.C. Lee, *Comparison of the different kinematic patterns during lateral bending between subjects with and without recurrent low back pain*. Clinical Biomechanics, 2016. **38**: p. 50-55.
20. Viggiani, D., et al., *The distribution of lumbar intervertebral angles in upright standing and extension is related to low back pain developed during standing*. Clinical Biomechanics, 2017. **49**: p. 85-90.
21. Izzo, R., et al., *Biomechanics of the spine. Part I: Spinal stability*. European journal of radiology, 2012. **82**.
22. Schmidler, C. *Lumbar Spine (Lower Back) Anatomy and Function*. 2015 [cited 2015 12 December]; Available from: <http://www.healthpages.org/anatomy-function/lumbar-spine-lower-back-structure-function/>.
23. Wu, G., et al., *ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion—part I: ankle, hip, and spine*. Journal of Biomechanics, 2002. **35**(4): p. 543-548.
24. Panjabi, M.M., M.H. Krag, and V.K. Goel, *A technique for measurement and description of three-dimensional six degree-of-freedom motion of a body joint with an application to the human spine*. Journal of Biomechanics, 1981. **14**(7): p. 447-460.
25. Dutton, M., *Orthopaedics for the physical therapist assistant*. 2011: Jones & Bartlett Publishers.
26. Watkins IV, R., et al., *Stability provided by the sternum and rib cage in the thoracic spine*. Spine, 2005. **30**(11): p. 1283-1286.
27. el-Khoury, G.Y. and C.G. Whitten, *Trauma to the upper thoracic spine: anatomy, biomechanics, and unique imaging features*. American Journal of Roentgenology, 1993. **160**(1): p. 95-102.
28. Rohlmann, A., et al., *Measurement of the number of lumbar spinal movements in the sagittal plane in a 24-hour period*. Eur Spine J, 2014. **23**(11): p. 2375-84.
29. Moroney, S.P., et al., *Load-displacement properties of lower cervical spine motion segments*. Journal of Biomechanics, 1988. **21**(9): p. 769-779.
30. Meakin, J.R., et al., *The intrinsic shape of the human lumbar spine in the supine, standing and sitting postures: characterization using an active shape model*. Journal of Anatomy, 2009. **215**(2): p. 206-211.
31. Roussouly, P., et al., *Classification of the normal variation in the sagittal alignment of the human lumbar spine and pelvis in the standing position*. Spine, 2005. **30**(3): p. 346-353.
32. Sullivan, M.S., C. Dickinson, and J. Troup, *The influence of age and gender on lumbar spine sagittal plane range of motion. A study of 1126 healthy subjects*. Spine, 1994. **19**(6): p. 682-686.
33. Wong, K.W., et al., *The flexion–extension profile of lumbar spine in 100 healthy volunteers*. Spine, 2004. **29**(15): p. 1636-1641.
34. Troke, M., et al., *A normative database of lumbar spine ranges of motion*. Man Ther, 2005. **10**(3): p. 198-206.
35. Milosavljevic, S., P.D. Milburn, and B.W. Knox, *The influence of occupation on lumbar sagittal motion and posture*. Ergonomics, 2005. **48**(6): p. 657-667.

36. Kuo, Y.-L., E.A. Tully, and M.P. Galea, *Skin movement errors in measurement of sagittal lumbar and hip angles in young and elderly subjects*. Gait & Posture, 2008. **27**(2): p. 264-270.
37. Hajibozorgi, M. and N. Arjmand, *Sagittal range of motion of the thoracic spine using inertial tracking device and effect of measurement errors on model predictions*. Journal of Biomechanics, 2016. **49**(6): p. 913-918.
38. Ignasiak, D., A. Rüeger, and S.J. Ferguson, *Multi-segmental thoracic spine kinematics measured dynamically in the young and elderly during flexion*. Human Movement Science, 2017. **54**: p. 230-239.
39. Zander, T., M. Bashkuev, and H. Schmidt, *Are there characteristic motion patterns in the lumbar spine during flexion?* Journal of Biomechanics, 2018. **70**: p. 77-81.
40. Narimani, M. and N. Arjmand, *Three-dimensional primary and coupled range of motions and movement coordination of the pelvis, lumbar and thoracic spine in standing posture using inertial tracking device*. Journal of Biomechanics, 2018. **69**: p. 169-174.
41. Kuhlman, K.A., *Cervical range of motion in the elderly*. Archives of physical medicine and rehabilitation, 1993. **74**(10): p. 1071-1079.
42. Dvořák, J., et al., *Normal motion of the lumbar spine as related to age and gender*. European Spine Journal, 1995. **4**(1): p. 18-23.
43. Intolo, P., et al., *The effect of age on lumbar range of motion: a systematic review*. Manual therapy, 2009. **14**(6): p. 596-604.
44. Vazirian, M., et al., *Lumbar contribution to the trunk forward bending and backward return; age-related differences*. Ergonomics, 2017. **60**(7): p. 967-976.
45. Muriuki, M.G., et al., *Effects of motion segment level, Pfirrmann intervertebral disc degeneration grade and gender on lumbar spine kinematics*. Journal of Orthopaedic Research, 2016. **34**(8): p. 1389-1398.
46. Kienbacher, T., et al., *Age and gender related neuromuscular changes in trunk flexion-extension*. Journal of neuroengineering and rehabilitation, 2015. **12**(1): p. 3.
47. Frey, M., R. Greene, and D. De Carvalho, *What is the best way to collect maximum forward lumbar spine flexion values for normalizing posture to range of motion?* Journal of Biomechanics, 2020. **103**: p. 109706.
48. Nissan, M. and I. Gilad, *Dimensions of human lumbar vertebrae in the sagittal plane*. Journal of Biomechanics, 1986. **19**(9): p. 753-758.
49. Hutton, W.C., B.M. Cyron, and J.R. Stott, *The compressive strength of lumbar vertebrae*. Journal of Anatomy, 1979. **129**(Pt 4): p. 753-758.
50. Smith, A., P. O'sullivan, and L. Straker, *Classification of sagittal thoracolumbo-pelvic alignment of the adolescent spine in standing and its relationship to low back pain*. Spine, 2008. **33**(19): p. 2101-2107.
51. Kendall, F., et al., *Testing and function with Posture and Pain*, in: 2005, Lippincoll Williams & Wilkins.
52. O'Sullivan, P., *Clinical instability' of the lumbar spine: its pathological basis, diagnosis and conservative management*. Grieve's Modern Manual Therapy-The Vertebral Column, 2004.
53. Sorensen, C.J., et al., *Is lumbar lordosis related to low back pain development during prolonged standing?* Manual Therapy, 2015. **20**(4): p. 553-557.

54. Quek, J., et al., *Effects of thoracic kyphosis and forward head posture on cervical range of motion in older adults*. Manual Therapy, 2013. **18**(1): p. 65-71.
55. Barrett, E., et al., *Is thoracic spine posture associated with shoulder pain, range of motion and function? A systematic review*. Manual Therapy, 2016. **26**: p. 38-46.
56. Goh, S., et al., *The relative influence of vertebral body and intervertebral disc shape on thoracic kyphosis*. Clinical Biomechanics, 1999. **14**(7): p. 439-448.
57. Baaklini, E., et al., *High-heeled walking decreases lumbar lordosis*. Gait & Posture, 2017. **55**: p. 12-14.
58. Sforza, C., et al., *Three-dimensional analysis of active head and cervical spine range of motion: effect of age in healthy male subjects*. Clinical Biomechanics, 2002. **17**(8): p. 611-614.
59. Chokshi, F.H., et al., *Radiographic Cobb Angle: A Feature of Congenital Lumbar Spine Stenosis*. Current Problems in Diagnostic Radiology, 2018.
60. Van Blommestein, A.S., et al., *Reliability of measuring thoracic kyphosis angle, lumbar lordosis angle and straight leg raise with an inclinometer*. Open Spine Journal, 2012.
61. Drzał-Grabiec, J., et al., *Comparison of Parameters Characterizing Lumbar Lordosis in Radiograph and Photogrammetric Examination of Adults*. Journal of Manipulative & Physiological Therapeutics, 2015. **38**(3): p. 225-231.
62. Chen, Y.-L., *Vertebral centroid measurement of lumbar lordosis compared with the Cobb technique*. Spine, 1999. **24**(17): p. 1786.
63. Chernukha, K.V., R.H. Daffner, and D.H. Reigel, *Lumbar lordosis measurement: a new method versus Cobb technique*. Spine, 1998. **23**(1): p. 74-79.
64. Polly Jr, D.W., et al., *Measurement of lumbar lordosis: evaluation of intraobserver, interobserver, and technique variability*. Spine, 1996. **21**(13): p. 1530-1535.
65. Dreischarf, M., et al., *Estimation of loads on human lumbar spine: A review of in vivo and computational model studies*. Journal of Biomechanics, 2016. **49**(6): p. 833-45.
66. Zhou, S.H., et al., *Geometrical dimensions of the lower lumbar vertebrae--analysis of data from digitised CT images*. Eur Spine J, 2000. **9**(3): p. 242-8.
67. Finestone, A.S., et al., *Assessing kyphosis with SpineScan: another attempt to reduce our dependence on radiography*. The Spine Journal, 2013. **13**(8): p. 926-931.
68. Thoumie, P., et al., *Effects of a lumbar support on spine posture and motion assessed by electrogoniometer and recording*. Clinical biomechanics, 1998. **13**(1): p. 18-26.
69. Snopek, A.M., *Fundamentals of Special Radiographic Procedures - E-Book*. 2013: Elsevier Health Sciences.
70. Campbell-Kyureghyan, N., et al., *The prediction of lumbar spine geometry: method development and validation*. Clinical Biomechanics, 2005. **20**(5): p. 455-464.
71. Mattmann, C., et al. *Recognizing upper body postures using textile strain sensors*. in *Wearable Computers, 2007 11th IEEE International Symposium on*. 2007. IEEE.

72. Bartalesi, R., et al. *Wearable monitoring of lumbar spine curvature by inertial and e-textile sensory fusion*. in *Engineering in Medicine and Biology Society (EMBC), 2010 Annual International Conference of the IEEE*. 2010. IEEE.
73. Walicka-Cupryś, K., J. Drzał-Grabiec, and M. Mrozkowiak, *Comparison of spine curvature parameters in the sagittal plane measured by the photogrammetric method and the inclinometer method*. 2013.
74. Kluszczyński, M., et al., *Prognostic value of measuring the angles of lumbar lordosis and thoracic kyphosis with the Saunders inclinometer in patients with low back pain*. Polish Annals of Medicine, 2017. **24**(1): p. 31-35.
75. Ernst, M.J., et al., *Extension and flexion in the upper cervical spine in neck pain patients*. Manual Therapy, 2015. **20**(4): p. 547-552.
76. Perrott, M.A., et al., *Comparison of lower limb and trunk kinematics between markerless and marker-based motion capture systems*. Gait & Posture, 2017. **52**: p. 57-61.
77. Lundberg, A., *On the use of bone and skin markers in kinematics research*. Human Movement Science, 1996. **15**(3): p. 411-422.
78. Chen, Y. and Y. Lee, *A non-invasive protocol for the determination of lumbosacral vertebral angle*. Clinical Biomechanics, 1997. **12**(3): p. 185-189.
79. Chen, Y.-L., *Predicting the vertebral inclination of the lumbar spine*. Ergonomics, 2000. **43**(6): p. 744-751.
80. Lee, Y.-H. and Y.-L. Chen, *Regressionally determined vertebral inclination angles of the lumbar spine in static lifts*. Clinical Biomechanics, 2000. **15**(9): p. 672-677.
81. Descarreaux, M., J.-S. Blouin, and N. Teasdale, *A non-invasive technique for measurement of cervical vertebral angle: report of a preliminary study*. European Spine Journal, 2003. **12**(3): p. 314-319.
82. Castro, A.P.G., et al., *Evaluation of spinal posture using Microsoft Kinect™: A preliminary case-study with 98 volunteers*. Porto Biomedical Journal, 2017. **2**(1): p. 18-22.
83. Hannink, E., et al., *Measurement of sagittal spine curvature: comparing the Kinect depth camera to the flexicurve and digital inclinometers in a clinical population*. Physiotherapy, 2020. **107**: p. e21.
84. Hannink, E., et al., *The validity of the kinect sensor for the measurement of sagittal spine curvature against the gold standard lateral spinal radiograph*. Physiotherapy, 2020. **107**: p. e28-e29.
85. Clark, R.A., et al., *Validity of the Microsoft Kinect for assessment of postural control*. Gait & posture, 2012. **36**(3): p. 372-377.
86. Destelle, F., et al. *Low-cost accurate skeleton tracking based on fusion of kinect and wearable inertial sensors*. in *2014 22nd European Signal Processing Conference (EUSIPCO)*. 2014.
87. Jana, A., *Kinect for Windows SDK Programming Guide*. 2012: Packt Pub.
88. Tsang, S.M.H., G.P.Y. Szeto, and R.Y.W. Lee, *Altered spinal kinematics and muscle recruitment pattern of the cervical and thoracic spine in people with chronic neck pain during functional task*. Journal of Electromyography and Kinesiology, 2014. **24**(1): p. 104-113.
89. Harlick, J.C., S. Milosavljevic, and P.D. Milburn, *Palpation identification of spinous processes in the lumbar spine*. Manual therapy, 2007. **12**(1): p. 56-62.
90. Merz, O., et al., *Validity of palpation techniques for the identification of the spinous process L5*. Manual therapy, 2013. **18**(4): p. 333-338.

91. Mieritz, R.M. and G.N. Kawchuk, *The accuracy of locating lumbar vertebrae when using palpation versus ultrasonography*. Journal of manipulative and physiological therapeutics, 2016. **39**(6): p. 387-392.
92. Shin, S., D.M. Yoon, and K.B. Yoon, *Identification of the correct cervical level by palpation of spinous processes*. Anesth Analg, 2011. **112**(5): p. 1232-5.
93. Kumar, S., *A study of spinal motion during lifting*. Irish Journal of Medical Science, 1974. **143**(1): p. 86-95.
94. Davis, P.R., J.D.G. Troup, and J.H. Burnard, *Movements of the thoracic and lumbar spine when lifting: A chrono-cyclophotographic study*. Journal of Anatomy, 1965. **99**(Pt 1): p. 13-26.
95. Allbrook, D., *Movements of the lumbar spinal column*. The Journal of bone and joint surgery. British volume, 1957. **39**(2): p. 339-345.
96. Panjabi, M., R. Brand, and A. White, *Mechanical properties of the human thoracic spine as shown by three-dimensional load-displacement curves*. The Journal of Bone & Joint Surgery, 1976. **58**(5): p. 642-652.
97. May, S. and D. Lomas, *Posture the Lumbar Spine and Back Pain*. International Encyclopedia of Rehabilitation, 2010.
98. Lind, B., et al., *Normal range of motion of the cervical spine*. Archives of Physical Medicine and Rehabilitation, 1989. **70**(9): p. 692-695.
99. Anderst, W.J., et al., *Cervical motion segment contributions to head motion during flexion\extension, lateral bending, and axial rotation*. The Spine Journal, 2015. **15**(12): p. 2538-2543.
100. Lord, M.J., et al., *Lumbar lordosis. Effects of sitting and standing*. Spine (Phila Pa 1976), 1997. **22**(21): p. 2571-4.
101. Claus, A.P., et al., *Thoracic and lumbar posture behaviour in sitting tasks and standing: Progressing the biomechanics from observations to measurements*. Applied Ergonomics, 2016. **53**, Part A: p. 161-168.
102. Wong, A.Y.L., et al., *Do different sitting postures affect spinal biomechanics of asymptomatic individuals?* Gait & Posture, 2019. **67**: p. 230-235.
103. A. Esola, M., et al., *Analysis of Lumbar Spine and Hip Motion During Forward Bending in Subjects With and Without a History of Low Back Pain*. Vol. 21. 1996. 71-8.
104. McKenzie, R. and S. May, *The lumbar spine: mechanical diagnosis and therapy*. Vol. 1. 2003: Orthopedic Physical Therapy.
105. Allbrook, D., *Movements of the lumbar spinal column*. Bone & Joint Journal, 1957. **39**(2): p. 339-345.
106. Wiles, P., *Movements of the Lumbar Vertebrae during Flexion and Extension:(Section of Orthopædics)*. Proceedings of the Royal Society of Medicine, 1935. **28**(5): p. 647.
107. Parkinson, S., et al., *Upper and lower lumbar segments move differently during sit-to-stand*. Manual Therapy, 2013. **18**(5): p. 390-394.
108. Tully, E.A., M.R. Fotoohabadi, and M.P. Galea, *Sagittal spine and lower limb movement during sit-to-stand in healthy young subjects*. Gait & Posture, 2005. **22**(4): p. 338-345.
109. O'Sullivan, P.B., et al., *The effect of different standing and sitting postures on trunk muscle activity in a pain-free population*. Spine, 2002. **27**(11): p. 1238-1244.
110. Crosbie, J., R. Vachalathiti, and R. Smith, *Patterns of spinal motion during walking*. Gait & Posture, 1997. **5**(1): p. 6-12.

111. Hu, P.-T., et al., *Effect of a cane on sit-to-stand transfer in subjects with hemiparesis*. American journal of physical medicine & rehabilitation, 2013. **92**(3): p. 191-202.
112. Carr, J.H., *Neurological rehabilitation : optimizing motor performance* / Janet H. Carr, Roberta B. Shepherd, ed. R.B. Shepherd. 2010, Edinburgh ; New York: Churchill Livingstone.
113. Shepherd, R.B. and A. Gentile, *Sit-to-stand: functional relationship between upper body and lower limb segments*. Human movement science, 1994. **13**(6): p. 817-840.
114. Chu, A.H.Y. and F.M. Moy, *Joint Association of Sitting Time and Physical Activity with Metabolic Risk Factors among Middle-Aged Malays in a Developing Country: A Cross-Sectional Study*. PLOS ONE, 2013. **8**(4): p. e61723.
115. Claus, A.P., et al., *Different ways to balance the spine in sitting: Muscle activity in specific postures differs between individuals with and without a history of back pain in sitting*. Clinical Biomechanics, 2018. **52**: p. 25-32.
116. Rohlmann, A., et al., *Comparison of intradiscal pressures and spinal fixator loads for different body positions and exercises*. Ergonomics, 2001. **44**(8): p. 781-794.
117. Chester, M.R., M.J. Rys, and S.A. Konz, *Leg swelling, comfort and fatigue when sitting, standing, and sit/standing*. International Journal of Industrial Ergonomics, 2002. **29**(5): p. 289-296.
118. Claus, A.P., et al., *Is 'ideal' sitting posture real?: Measurement of spinal curves in four sitting postures*. Manual Therapy, 2009. **14**(4): p. 404-408.
119. Granata, K.P. and A.H. Sanford, *Lumbar–pelvic coordination is influenced by lifting task parameters*. Spine, 2000. **25**(11): p. 1413-1418.
120. Tafazzol, A., et al., *Lumbopelvic rhythm during forward and backward sagittal trunk rotations: combined in vivo measurement with inertial tracking device and biomechanical modeling*. Clinical Biomechanics, 2014. **29**(1): p. 7-13.
121. Rathore, M., et al., *A focused review–thoracolumbar spine: anatomy, biomechanics and clinical significance*. Indian Journal of Clinical Anatomy and Physiology Vol, 2014. **1**(1).
122. Pollintine, P., et al., *Intervertebral disc degeneration can lead to “stress-shielding” of the anterior vertebral body: a cause of osteoporotic vertebral fracture?* Spine, 2004. **29**(7): p. 774-782.
123. Chen, Y.-L., et al., *Effect of individual flexibility and knee posture on the lumbar curvature and back muscle flexion-relaxation phenomenon*. International Journal of Industrial Ergonomics, 2018. **68**: p. 82-88.
124. Nordin, M., et al., *Spinal configuration during lifting*. Bulletin of the Hospital for Joint Diseases Orthopaedic Institute, 1985. **46**(1): p. 31-36.
125. Cobian, D.G., et al., *Active cervical and lumbar range of motion during performance of activities of daily living in healthy young adults*. Spine, 2013. **38**(20): p. 1754-1763.
126. Wang, S.-F., C.-C. Teng, and K.-H. Lin, *Measurement of cervical range of motion pattern during cyclic neck movement by an ultrasound-based motion system*. Manual Therapy, 2005. **10**(1): p. 68-72.
127. Alqhtani, R.S., et al., *Investigating the contribution of the upper and lower lumbar spine, relative to hip motion, in everyday tasks*. Manual Therapy, 2016. **21**: p. 268-273.

128. Alqhtani, R.S., et al., *Correlation of lumbar-hip kinematics between trunk flexion and other functional tasks*. Journal of manipulative and physiological therapeutics, 2015. **38**(6): p. 442-447.
129. Yoshioka, S., et al., *The minimum required muscle force for a sit-to-stand task*. Journal of Biomechanics, 2012. **45**(4): p. 699-705.
130. Widhalm, K., T. Stamm, and E.J. Hurkmans, *Coordination of the hip and lumbar spine during sit-to-stand in healthy subjects*. Gait & Posture, 2015. **42**: p. S50-S51.
131. Sibella, F., et al., *Biomechanical analysis of sit-to-stand movement in normal and obese subjects*. Clinical Biomechanics, 2003. **18**(8): p. 745-750.
132. Pourahmadi, M.R., et al., *Test-retest reliability of sit-to-stand and stand-to-sit analysis in people with and without chronic nonspecific low back pain*. Annals of Physical and Rehabilitation Medicine, 2018. **61**: p. e153.
133. Ippersiel, P., S. Robbins, and R. Preuss, *Movement variability in adults with low back pain during sit-to-stand-to-sit*. Clinical Biomechanics, 2018. **58**: p. 90-95.
134. Fotoohabadi, M.R., E. Tully, and M. Galea, *Sagittal kinematics of the hip-spine interaction during sit-to-stand in healthy subjects*. Journal of Biomechanics, 2006. **39**: p. S542.
135. Bohannon, R.W., et al., *Sit-to-stand test: Performance and determinants across the age-span*. Isokinetics and exercise science, 2010. **18**(4): p. 235-240.
136. Papi, E., A.M.J. Bull, and A.H. McGregor, *Spinal segments do not move together predictably during daily activities*. Gait & Posture, 2019. **67**: p. 277-283.
137. Vaughn, J.J. and R.M. Schwend, *Sitting sagittal balance is different from standing balance in children with scoliosis*. Journal of Pediatric Orthopaedics, 2014. **34**(2): p. 202-207.
138. Proper, K.I., et al., *Sedentary behaviors and health outcomes among adults: a systematic review of prospective studies*. American journal of preventive medicine, 2011. **40**(2): p. 174-182.
139. Korakakis, V., et al., *Repeated end range spinal movement while seated abolishes the proprioceptive deficit induced by prolonged flexed sitting posture. A study assessing the statistical and clinical significance of spinal position sense*. Musculoskeletal Science and Practice, 2017. **31**: p. 9-20.
140. Harrison, D.D., et al., *Sitting biomechanics Part I: Review of the Literature*. Journal of Manipulative and Physiological Therapeutics, 1999. **22**(9): p. 594-609.
141. Fon, G.T., M.J. Pitt, and A.C. Thies, *Thoracic kyphosis: range in normal subjects*. American Journal of Roentgenology, 1980. **134**(5): p. 979-983.
142. Nicholas, J.A. and P.D. Wilson, *3 Osteoporosis of the Aged Spine*. Clinical orthopaedics and related research, 1963. **26**: p. 19-33.
143. Rowe, C.R. and C. Sorbie, *4 Fractures of the Spine in the Aged*. Clinical orthopaedics and related research, 1963. **26**: p. 34-49.
144. Liu, J., *A two-dimensional human spine simulation and three-dimensional spine model construction*. 2001, University of Florida.
145. Poortmans, V., et al., *Assessment of lumbar spine height following sustained lumbar extension posture: comparison between musculoskeletal ultrasonography and stadiometry*. Journal of Manipulative & Physiological Therapeutics, 2016. **39**(8): p. 586-593.

146. Rodríguez, S., et al., *Estimating height from the first and second cervical vertebrae in a Spanish population*. Legal Medicine, 2016. **19**: p. 88-92.
147. Pape, J.L., et al., *Increased spinal height using propped slouched sitting postures: Innovative ways to rehydrate intervertebral discs*. Applied ergonomics, 2018. **66**: p. 9-17.
148. Mörl, F., H. Wagner, and R. Blickhan, *Lumbar spine intersegmental motion analysis during lifting*. Pathophysiology, 2005. **12**(4): p. 295-302.
149. Yanagisawa, S., et al., *Relation among the knee, sagittal spinal alignment, and the spinal range of motion: Investigation in local medical check-ups using the SpinalMouse*. Asia-Pacific Journal of Sports Medicine, Arthroscopy, Rehabilitation and Technology, 2015. **2**(2): p. 68-71.
150. Wang, H., et al., *The effect of active sitting on trunk motion*. Journal of Sport and Health Science, 2014. **3**(4): p. 333-337.
151. Lee, R.Y.W. and T.K.T. Wong, *Relationship between the movements of the lumbar spine and hip*. Human Movement Science, 2002. **21**(4): p. 481-494.
152. Walsh, R. *What is an orthosis?* 2017 7 December 2017; Available from: <https://www.abclawcenters.com/blog/2017/12/07/what-is-an-orthosis/>.
153. Lee, W.-K., *13 - Spinal Orthoses*, in *Braddom's Rehabilitation Care: A Clinical Handbook*, D.X. Cifu and H.L. Lew, Editors. 2018, Elsevier. p. 85-91.e2.
154. Rao, S., et al., *An Overview of Taguchi Method: Evolution, Concept and Interdisciplinary Applications*. International Journal of Scientific & Engineering Research, 2013. **4**(10): p. 621-626.
155. Curfs, I., et al., *Evaluating the Immobilization Effect of Spinal Orthoses Using Sensor-Based Motion Analysis*. JPO: Journal of Prosthetics and Orthotics, 2016. **28**(1): p. 23-29.
156. Lang, G., et al., *Efficacy of Thoracolumbar Bracing in Spinal Immobilization: Precise Assessment of Gross, Intersegmental, and Segmental Spinal Motion Restriction by a 3D Kinematic System*. World Neurosurgery, 2018. **116**: p. e128-e146.
157. Azadinia, F., et al., *The effects of two spinal orthoses on balance in elderly people with thoracic kyphosis*. Prosthetics And Orthotics International, 2013. **37**(5): p. 404-410.
158. Y. Dionyssiotis, I.A.D., D. Economopoulos, G.P. Lyritis, *Rehabilitation after falls and fractures*. 2008: p. 244 - 250.
159. Seymour, R., *Prosthetics and Orthotics, Lower limb and Spinal*. 2002, 351 West Camden Street, Baltimore, Maryland 21201-2436 USA: Lippincott Williams & Wilkins. 485.
160. Picchiotti, M.T., et al., *Impact of two postural assist exoskeletons on biomechanical loading of the lumbar spine*. Applied Ergonomics, 2019. **75**: p. 1-7.
161. Hofmann, U.K., et al., *The Effects of Hip and Spine Orthoses on Braking Parameters: A Simulated Study With Healthy Subjects*. PM&R, 2016. **8**(1): p. 35-44.
162. Baltrusch, S., et al., *SPEXOR passive spinal exoskeleton decreases metabolic cost during symmetric repetitive lifting*. European Journal of Applied Physiology, 2020. **120**(2): p. 401-412.
163. De Rijcke, L., et al., *SPEXOR: Towards a passive spinal exoskeleton*. 2017. p. 325-329.

164. Baltrusch, S., et al., *Testing an Exoskeleton That Helps Workers With Low-Back Pain: Less Discomfort with the Passive SPEXOR Trunk Device*. IEEE Robotics & Automation Magazine, 2019.
165. Pfeifer, M., B. Begerow, and H.W. Minne, *Effects of a new spinal orthosis on posture, trunk strength, and quality of life in women with postmenopausal osteoporosis: a randomized trial*. American journal of physical medicine & rehabilitation, 2004. **83**(3): p. 177-186.
166. Vogt, L., et al., *Postural correction by osteoporosis orthosis (Osteo-med): a randomized, placebo-controlled trial*. Prosthetics and orthotics international, 2008. **32**(1): p. 103-110.
167. Wong, M.S., et al., *The effect of rigid versus flexible spinal orthosis on the clinical efficacy and acceptance of the patients with adolescent idiopathic scoliosis*. Spine, 2008. **33**(12): p. 1360-1365.
168. Pfeifer, M., et al., *Effects of two newly developed spinal orthoses on trunk muscle strength, posture, and quality-of-life in women with postmenopausal osteoporosis: a randomized trial*. American journal of physical medicine & rehabilitation, 2011. **90**(10): p. 805-815.
169. Valentin, G.H., L.N. Pedersen, and T. Maribo, *Wearing an active spinal orthosis improves back extensor strength in women with osteoporotic vertebral fractures*. Prosthetics and Orthotics International, 2014. **38**(3): p. 232-238.
170. Näf, M.B., et al., *Passive back support exoskeleton improves range of motion using flexible beams*. Frontiers in Robotics and AI, 2018. **5**: p. 72.
171. Program, C.H.B.R., *Analysis of Assembly Bill 2012-Amended: Orthotic and Prosthetic Devices*. 2006: eScholarship, University of California.
172. Naruse, K., et al., *Design of wearable power-assist device for lower back support*. Journal of Robotics and Mechatronics, 2004. **16**(5): p. 489-496.
173. Naruse, K., et al. *Development of wearable exoskeleton power assist system for lower back support*. in *Intelligent Robots and Systems, 2003.(IROS 2003). Proceedings. 2003 IEEE/RSJ International Conference on*. 2003. IEEE.
174. Johnson, D.D., J.A. Ashton-Miller, and A.J. Shih, *Active Spinal Orthosis to Reduce Lumbar Postural Muscle Activity in Flexed Postures*. JPO: Journal of Prosthetics and Orthotics, 2016. **28**(3): p. 109-113.
175. Zhang, H., A. Kadrolkar, and F.C. Sup, *Design and preliminary evaluation of a passive spine exoskeleton*. Journal of Medical Devices, 2016. **10**(1).
176. Toxiri, S., et al. *A wearable device for reducing spinal loads during lifting tasks: Biomechanics and design concepts*. in *2015 IEEE International Conference on Robotics and Biomimetics (ROBIO)*. 2015. IEEE.
177. Ballesteros-Escamilla, M., et al., *Hybrid position/force output feedback second-order sliding mode control for a prototype of an active orthosis used in back-assisted mobilization*. Medical & biological engineering & computing, 2019. **57**(9): p. 1843-1860.
178. Taghi, M.K., *The Physiological Benefits and Problems Associated With Using Standing and Walking Orthoses in Individuals With Spinal Cord Injury—A Meta-analytic Review*. Journal of Orthopaedics, Trauma and Rehabilitation, 2012. **16**(1): p. 37-40.
179. Rohlmann, A., et al., *Effect of an orthosis on the loads acting on a vertebral body replacement*. Clinical Biomechanics, 2013. **28**(5): p. 490-494.
180. Colobățiu, L. and S. Mirel, *The efficacy of modern spinal orthoses and hip protectors used in the management of osteoporosis: a review*. Balneo Research Journal, 2016. **7**(2): p. 60-65.

181. Rimmer, J.H., D. Braddock, and K.H. Pitetti, *Research on physical activity and disability: an emerging national priority*. Medicine & Science in Sports & Exercise, 1996. **28**(11): p. 1366-1372.
182. Agabegi, S.S., F.A. Asghar, and H.N. Herkowitz, *Spinal orthoses*. JAAOS-Journal of the American Academy of Orthopaedic Surgeons, 2010. **18**(11): p. 657-667.
183. Naruse, K., et al., *Design of wearable power-assist device for lower back support*. Journal of Robotics and Mechatronics, 2004. **16**: p. 489-496.
184. Naruse, K., S. Kawai, and T. Kukichi. *Three-dimensional lifting-up motion analysis for wearable power assist device of lower back support*. in *Intelligent Robots and Systems, 2005.(IROS 2005). 2005 IEEE/RSJ International Conference on*. 2005. IEEE.
185. Tanaka, T., et al. *Smart suit: Soft power suit with semi-active assist mechanism-prototype for supporting waist and knee joint*. in *Control, Automation and Systems, 2008. ICCAS 2008. International Conference on*. 2008. IEEE.
186. Imamura, Y., et al. *Motion-based design of elastic belts for passive assistive device using musculoskeletal model*. in *Robotics and Biomimetics (ROBIO), 2011 IEEE International Conference on*. 2011. IEEE.
187. Imamura, Y., et al., *Motion-based-design of elastic material for passive assistive device using musculoskeletal model*. Journal of Robotics and Mechatronics, 2011. **23**(6): p. 978.
188. Imamura, Y., et al. *Verification of assistive effect generated by passive power-assist device using humanoid robot*. in *System Integration (SII), 2014 IEEE/SICE International Symposium on*. 2014. IEEE.
189. Frost, D.M., M. Abdoli-E, and J.M. Stevenson, *PLAD (personal lift assistive device) stiffness affects the lumbar flexion/extension moment and the posterior chain EMG during symmetrical lifting tasks*. Journal of Electromyography and Kinesiology, 2009. **19**(6): p. e403-e412.
190. Sadler, E.M., R.B. Graham, and J.M. Stevenson, *The personal lift-assist device and lifting technique: a principal component analysis*. Ergonomics, 2011. **54**(4): p. 392-402.
191. Wehner, M., D. Rempel, and H. Kazerooni. *Lower extremity exoskeleton reduces back forces in lifting*. in *ASME 2009 dynamic systems and control conference*. 2009. American Society of Mechanical Engineers Digital Collection.
192. Li, X., et al., *Design of wearable power assist wear for low back support using pneumatic actuators*. International Journal of Automation Technology (IJAT), 2013. **7**(2): p. 228-236.
193. Ikeya, Y. and T. Nakakuki. *A study on a waist power assist suit for a caregiver to prevent lower back pain*. in *Control, Automation and Systems (ICCAS), 2013 13th International Conference on*. 2013. IEEE.
194. Muramatsu, Y. and H. Kobayashi, *Assessment of local muscle fatigue by NIRS-development and evaluation of muscle suit*. ROBOMECH Journal, 2014. **1**(1): p. 19.
195. Muramatsu, Y., H. Umehara, and H. Kobayashi. *Improvement and quantitative performance estimation of the back support muscle suit*. in *2013 35th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC)*. 2013. IEEE.
196. Chen, B., et al., *A wearable exoskeleton suit for motion assistance to paralysed patients*. Journal of Orthopaedic Translation, 2017. **11**: p. 7-18.

197. Inose, H., et al. *Semi-endoskeleton-type waist assist AB-wear suit equipped with compressive force reduction mechanism*. in *2017 IEEE International Conference on Robotics and Automation (ICRA)*. 2017. IEEE.
198. Koopman, A.S., et al., *The effect of control strategies for an active back-support exoskeleton on spine loading and kinematics during lifting*. *Journal of Biomechanics*, 2019. **91**: p. 14-22.
199. Yin, P., et al., *Effects of wearable power assist device on low back fatigue during repetitive lifting tasks*. *Clinical Biomechanics*, 2019. **70**: p. 59-65.
200. Sun, P.S., et al., *Inertial sensors-based torso motion mode recognition for an active postural support brace*. *Advanced Robotics*, 2020. **34**(1): p. 57-67.
201. Sun, P.S., et al., *Upper-Body Motion Mode Recognition Based on IMUs for a Dynamic Spine Brace*. *2018 IEEE International Conference on Cyborg and Bionic Systems (CBS)*, 2018: p. 167-170.
202. Koo, T.K. and M.Y. Li, *A Guideline of Selecting and Reporting Intraclass Correlation Coefficients for Reliability Research*. *Journal of chiropractic medicine*, 2016. **15**(2): p. 155-163.
203. Field, A.P., *Intraclass correlation*. Wiley StatsRef: Statistics Reference Online, 2014.
204. Chen, C.-C. and H.X. Barnhart, *Assessing agreement with intraclass correlation coefficient and concordance correlation coefficient for data with repeated measures*. *Computational Statistics & Data Analysis*, 2013. **60**: p. 132-145.
205. Shrout, P.E. and J.L. Fleiss, *Intraclass correlations: uses in assessing rater reliability*. *Psychological bulletin*, 1979. **86**(2): p. 420.
206. Edwards, A.W.F., *Chapter 67 - R.A. Fischer, statistical methods for research workers, first edition (1925)*, in *Landmark Writings in Western Mathematics 1640-1940*, I. Grattan-Guinness, et al., Editors. 2005, Elsevier Science: Amsterdam. p. 856-870.
207. McGraw, K.O. and S.P. Wong, *Forming inferences about some intraclass correlation coefficients*. *Psychological methods*, 1996. **1**(1): p. 30.
208. Seerden, S.F.L., et al., *Multi-segment spine and hip kinematics in asymptomatic individuals during standardized return from forward bending versus functional box lifting*. *Journal of Electromyography and Kinesiology*, 2019. **49**: p. 102352.
209. Ahn, J.-S., et al., *Comparison of Clinical Results according to the Complications after or during Open Door Laminoplasty Surgery for Cervical Myelopathy*. *Journal of the Korean Orthopaedic Association*, 2013. **48**(5): p. 375-381.
210. Roghani, T., et al., *The Reliability of Standing Sagittal Measurements of Spinal Curvature and Range of Motion in Older Women With and Without Hyperkyphosis Using a Skin-Surface Device*. *Journal of Manipulative and Physiological Therapeutics*, 2017. **40**(9): p. 685-691.
211. Tokgoz, N., et al., *Are Spinal or Paraspinal Anatomic Markers Helpful for Vertebra Numbering and Diagnosing Lumbosacral Transitional Vertebrae?* Vol. 15. 2014. 258-266.
212. McKinnis, L.N., *Fundamentals of Musculoskeletal Imaging*. 2013: F. A. Davis Company.
213. Kim, K.H., et al., *Changes in Spinal Canal Diameter and Vertebral Body Height with Age*. *Yonsei Med J*, 2013. **54**(6): p. 1498-1504.

214. Weber, U., et al., *Whole body MR imaging in ankylosing spondylitis: a descriptive pilot study in patients with suspected early and active confirmed ankylosing spondylitis*. BMC Musculoskeletal Disorders, 2007. **8**(1): p. 20.
215. Singer, K.P., et al., *Computer-assisted curvature assessment and Cobb angle determination of the thoracic kyphosis. An in vivo and in vitro comparison*. Spine (Phila Pa 1976), 1994. **19**(12): p. 1381-4.
216. Shum, G.L., J. Crosbie, and R.Y. Lee, *Effect of low back pain on the kinematics and joint coordination of the lumbar spine and hip during sit-to-stand and stand-to-sit*. Spine, 2005. **30**(17): p. 1998-2004.
217. Johnson, M.B., et al., *Multi-segmental torso coordination during the transition from sitting to standing*. Clinical Biomechanics, 2010. **25**(3): p. 199-205.
218. Bible, J.E., et al., *Normal functional range of motion of the cervical spine during 15 activities of daily living*. Clinical Spine Surgery, 2010. **23**(1): p. 15-21.
219. Fotoohabadi, M.R., E.A. Tully, and M.P. Galea, *Kinematics of Rising From a Chair: Image-Based Analysis of the Sagittal Hip-Spine Movement Pattern in Elderly People Who Are Healthy*. Physical Therapy, 2010. **90**(4): p. 561-571.
220. Ignasiak, D., et al., *Thoracolumbar spine loading associated with kinematics of the young and the elderly during activities of daily living*. Journal of Biomechanics, 2018. **70**: p. 175-184.
221. Wang, X., et al., *Repeatability of Cervical Joint Flexion and Extension Within and Between Days*. Journal of Manipulative and Physiological Therapeutics, 2018. **41**(1): p. 10-18.
222. Mousavi-Khatir, R., et al., *Effect of static neck flexion in cervical flexion-relaxation phenomenon in healthy males and females*. Journal of Bodywork and Movement Therapies, 2016. **20**(2): p. 235-242.
223. Cho, I.Y., et al., *The Effect of Standing and Different Sitting Positions on Lumbar Lordosis: Radiographic Study of 30 Healthy Volunteers*. Asian Spine J, 2015. **9**(5): p. 762-769.
224. Heneghan, N.R. and A. Rushton, *Understanding why the thoracic region is the 'Cinderella' region of the spine*. Manual Therapy, 2016. **21**: p. 274-276.
225. Hey, H.W.D., et al., *How the spine differs in standing and in sitting—important considerations for correction of spinal deformity*. The Spine Journal, 2017. **17**(6): p. 799-806.
226. Alessa, F. and X. Ning, *Changes of lumbar posture and tissue loading during static trunk bending*. Human Movement Science, 2018. **57**: p. 59-68.
227. du Rose, A., A. Breen, and A. Breen, *Relationships between muscle electrical activity and the control of inter-vertebral motion during a forward bending task*. Journal of Electromyography and Kinesiology, 2018. **43**: p. 48-54.
228. Panjabi, M.M., *The stabilizing system of the spine. Part I. Function, dysfunction, adaptation, and enhancement*. J Spinal Disord, 1992. **5**(4): p. 383-9; discussion 397.
229. Beaudette, S.M., et al., *The effect of head and gaze orientation on spine kinematics during forward flexion*. Human Movement Science, 2020. **70**: p. 102590.
230. Mörl, F., I. Bradl, and R. Blickhan, *Differences in lumbar spine motion pattern of weight-lifters and normal subjects performing freestyle lifting trials*. Journal of Biomechanics, 2006. **39**: p. S172.
231. Damm, P., et al., *In vivo hip and lumbar spine implant loads during activities in forward bent postures*. Journal of Biomechanics, 2020. **102**: p. 109517.

232. Mitnitski, A.B., et al., *Coordination between the lumbar spine lordosis and trunk angle during weight lifting*. Clinical Biomechanics, 1998. **13**(2): p. 121-127.
233. Swie, Y.W., K. Sakamoto, and Y. Shimizu, *Chaotic analysis of electromyography signal at low back and lower limb muscles during forward bending posture*. Electromyography and clinical neurophysiology, 2005. **45**(6): p. 329-342.
234. Gagnon, D., C. Larivière, and P. Loisel, *Comparative ability of EMG, optimization, and hybrid modelling approaches to predict trunk muscle forces and lumbar spine loading during dynamic sagittal plane lifting*. Clinical Biomechanics, 2001. **16**(5): p. 359-372.
235. Larivière, C., D. Gagnon, and P. Loisel, *The comparison of trunk muscles EMG activation between subjects with and without chronic low back pain during flexion–extension and lateral bending tasks*. Journal of Electromyography and Kinesiology, 2000. **10**(2): p. 79-91.
236. Granata, K.P. and W.S. Marras, *An EMG-assisted model of loads on the lumbar spine during asymmetric trunk extensions*. Journal of Biomechanics, 1993. **26**(12): p. 1429-1438.
237. El-Rich, M., A. Shirazi-Adl, and N. Arjmand, *Muscle Activity, Internal Loads, and Stability of the Human Spine in Standing Postures: Combined Model and In Vivo Studies*. Spine, 2004. **29**(23): p. 2633-2642.
238. Hashemirad, F., et al., *The relationship between flexibility and EMG activity pattern of the erector spinae muscles during trunk flexion–extension*. Journal of Electromyography and Kinesiology, 2009. **19**(5): p. 746-753.
239. Kingma, I., G.S. Faber, and J.H. van Dieën, *Supporting the upper body with the hand on the thigh reduces back loading during lifting*. Journal of Biomechanics, 2016. **49**(6): p. 881-889.
240. Ekstrom, R.A., R.W. Osborn, and P.L. Hauer, *Surface Electromyographic Analysis of the Low Back Muscles During Rehabilitation Exercises*. Journal of Orthopaedic & Sports Physical Therapy, 2008. **38**(12): p. 736-745.
241. Takahashi, I., et al., *Mechanical load of the lumbar spine during forward bending motion of the trunk—a biomechanical study*. Spine, 2006. **31**(1): p. 18-23.
242. Kent, R., *Quality Management in Plastics Processing*. 2016: Elsevier.
243. Pham, H., M. Le, and V. Mai, *A Novel Multi-Axis Compliant Prosthetic Ankle Foot to Support the Rehabilitation of Amputees*. in *2016 3rd International Conference on Green Technology and Sustainable Development (GTSD)*. 2016.
244. Karimi, M., et al., *An evaluation of the efficiency of endpoint control on the correction of scoliotic curve with brace: A case study*. Acta of bioengineering and biomechanics, 2019. **21**(2).
245. Perie, D., J.S. De Gauzy, and M. Hobatho, *Biomechanical evaluation of Cheneau-Toulouse-Munster brace in the treatment of scoliosis using optimisation approach and finite element method*. Medical and Biological Engineering and Computing, 2002. **40**(3): p. 296-301.
246. Mafi, M. and A.M.K. Niyaval, *Brace Simulation for Patients with Scoliosis: Biomechanical Comparison with Three Different Polymeric Materials*. The Bangkok Medical Journal, 2019. **15**(2): p. 161-161.
247. Wang, L., et al., *Anisotropic design of a multilayered biological exoskeleton*. Journal of Materials Research, 2009. **24**(12): p. 3477-3494.

248. Phan, T.-V., H.-T. Pham, and C.-N. Truong, *Design and Analysis of a Compliant Constant-Torque Mechanism for Rehabilitation Devices*, in *Advanced Materials*. 2020, Springer. p. 541-549.
249. Karimkhani, B., S.A. Mousavi, and M. Sattari, *Design and Construction of Electromechanical Wrist Hand Orthosis with a Functional Interface*. *Journal of Modern Processes in Manufacturing and Production*, 2018. 7(4): p. 57-70.
250. Xu, Y. *Design and simulation of a soft robotic device for wrist rehabilitation*. in *2020 3rd International Conference on Advanced Electronic Materials, Computers and Software Engineering (AEMCSE)*. 2020. IEEE.
251. Afripin, M.A.A., *Damage-Mechanics Based Model for Reliability Assessment of Through-Silicon Via Interconnects*, in *School of Mechanical Engineering, Faculty of Engineering* 2020, Universiti Teknologi Malaysia. p. 153.
252. Lim, T., et al., *Distribution of body weight, height and body mass index in a national sample of Malaysian adults*. *Medical Journal of Malaysia*, 2000. 55(1): p. 108-128.

Appendix A Product Design Specifications, PDS

PDS conducted for the proposed spinal orthosis

	Competitive Best	My Design
Aesthetic	Spinomed Modern strap design California Cruciform Extension Orthosis low-profile design fits under clothing	Modern harness design. Provide an attractive combination of colors for the outer surface of the spinal orthosis.
Quality and Reliability	Spinomed. Clinically proven reliability results : Strengthen back extensors by 73% Strengthen abdominal flexors by 58% Reduce the angle of kyphosis by 11% Reduce daily pain by 38% Reduce body sway by 25% Increase vital capacity by 7%	Provide a low-maintenance product with high reliability.
User	Spinomed® IV Ergonomically shaped shoulder strap that increases wearing comfort Active breathing Ergonomically pre-formed shoulder straps California Catalina Custom Fit 631/637 Lightweight and comfortable	Controllable spinal orthosis. Ergonomic structure with the 3-point pressure system. Lightweight and comfortable harness design. A high degree of wearing comfort.
Life in Service Material and workmanship. Also if needed, include product maintenance.	Bledshoe Spinal orthosis braces (90 days for material & workmanship. Lifetime warranty for pulley system) Ottobok (US Warranty Duration) Orthotic Components for use in custom fabrication (12 months). Orthotic Pre-Fabricated Braces. (6 months) California Catalina Custom Fit 627 Adjustable waist size, no cut. *look for the catalog	The target lifespan is for 10 years. Only use simple mechanisms and maintenance work to reduce the complexity if problems occur. Users can repair or replace the part easily by themselves.
Material	California® Ventura™ Spinal Brace Removable rigid plastic components Spinomed® IV The panels are covered in a breathable, moisture-wicking fabric	Use of soft material and fabric to ensure a high degree of comfort (harness). The material used for the joint mechanism may make up of strong steel/aluminum or any metal structure.
Safety features	California Catalina Custom Fit 631/637 Simple to don and doff	Provide with emergency system breakdown controller. Easy to don or doff.
Cost comprising	MEDI USA Spinomed IV Osteoporosis Spinal Orthosis RM 1387.05 (\$325) Medi USA 4C Flex Hyper Extension Spinal Orthosis RM 1066.92 (\$249.99)	Target price - Ranged between RM500-RM1000

Continue...

Standard & regulation	Local Coverage Determination (LCD): SPINAL ORTHOSES: TLSO and LSO (L11459) & DME Medical Policies Cover up to all spinal orthosis. A thoracic-lumbar-sacral orthosis (L0450-L0492), lumbar orthosis (L0625-L0627), or lumbar-sacral orthosis (L0628-L0640) is covered when it is ordered for one of the following indications: To reduce pain by restricting the mobility of the trunk; or To facilitate healing following an injury to the spine or related soft tissues; or To facilitate healing following a surgical procedure on the spine or related soft tissue; or To otherwise support weak SPINAL muscles and/or a deformed spine. If a SPINAL orthosis is provided and the coverage criteria are not met, the item will be denied as not medically necessary. Thoracic-lumbar-sacral orthoses (TLSO) described by codes L0450 - L0492, lumbar orthoses (LO) described by codes, L0625 - L0627, and lumbar-sacral orthoses (LSO) described by codes L0628 - L0640, all have the following characteristics: Used to immobilize the specified areas of the spine. Intimate fit and generally designed to be worn under clothing. Not specifically designed for patients in wheelchairs.
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Appendix B Morphological Chart

The morphological chart for the initial spinal orthosis presented is separated into eight sections: actuator, sensor, joint mechanism, joint mechanism material, harness design, aesthetic value, power system, and an emergency mechanism. The mechanism's detailed drawing may be seen in Appendix D.

Details on morphological chart of the proposed multi-joint spinal orthosis

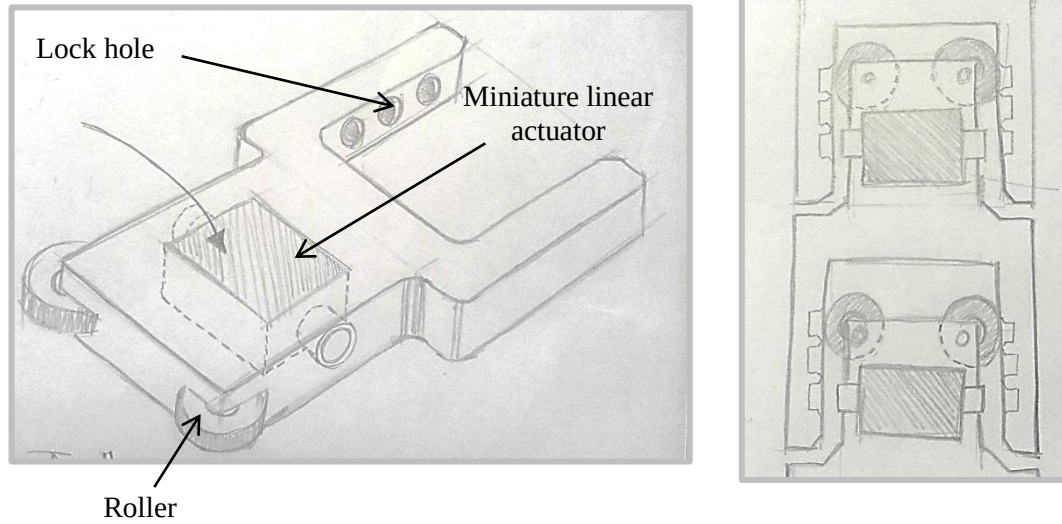
Concept	A	B	C	D
Component				
Actuator	Piezo motor 4	Compression/ memory springs	Miniature linear actuator 1	Solenoid 2
Sensor	EMG sensors	Gyro and Accelerometer sensors	Bend sensor 3	Pressure sensors
Joint mechanism	A	B	C	D
Material for joint mechanism	Plastic	Aluminium	Steel	
Harness design	A	B		
Aesthetic value	Attractive combination of colour for outer surface	Cushion frame		
Power system	DC Power Supply	AC Power Supply	Rechargeable Battery	
Emergency mechanism	Stop button on the machine at exoskeleton	Switch on controller	PLC	

Appendix C Summary of different concepts proposed

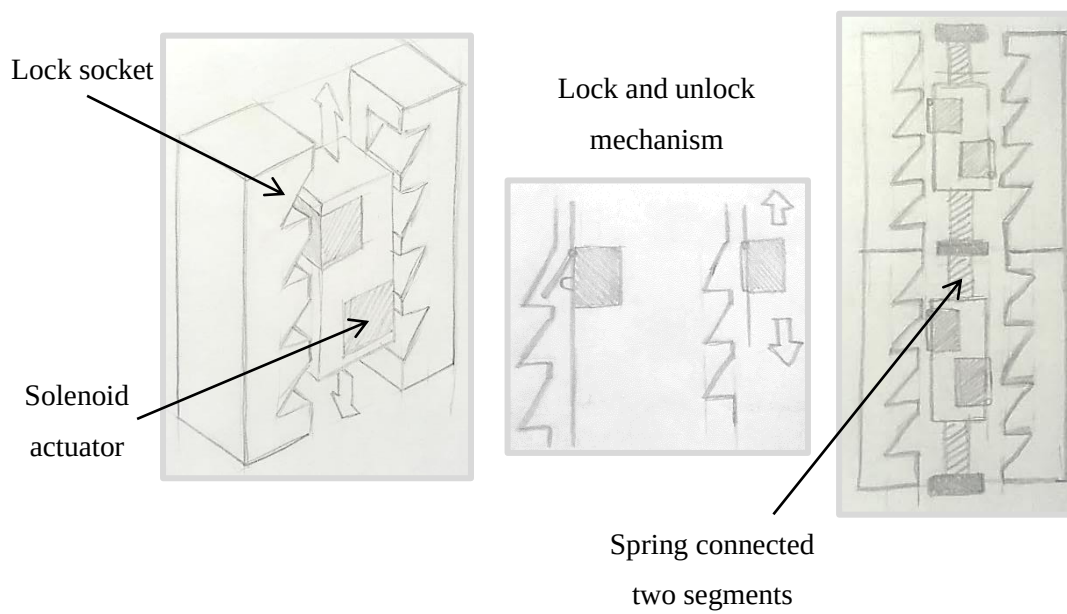
Concept	1	2	3	4
Actuator	Miniature linear actuator	Solenoid	Miniature linear actuator	Piezomotor
Sensor	Gyro and accelerometer sensors	bend sensor	bend sensor	bend sensor
Joint mechanism	A	B	C	D
Material	Plastic	Aluminium	Aluminium	Steel
Harness design	A	A	B	B
Power system	DC Power Supply	Rechargeable Battery	Rechargeable Battery	Rechargeable Battery
Emergency mechanism	Switch on controller	Stop button on the machine at exoskeleton	PLC	Switch on controller

Appendix D Details of the joint mechanism (with actuator)

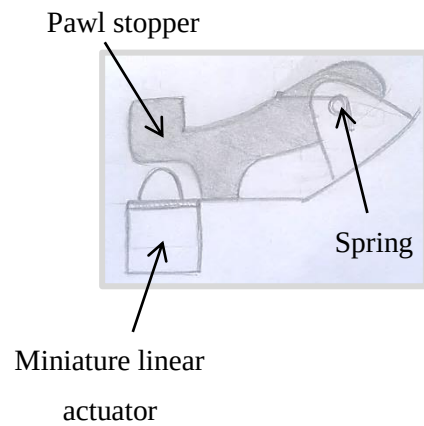
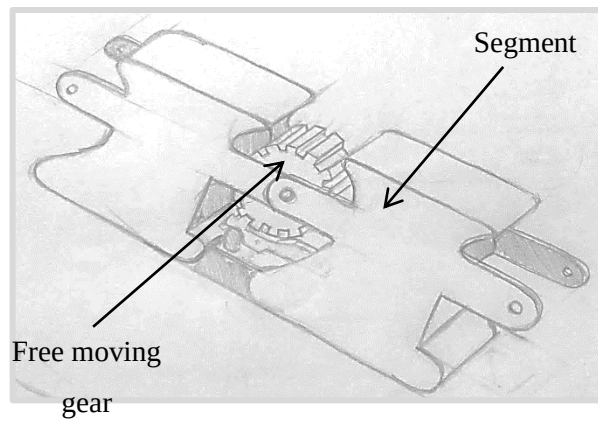
Concept 1



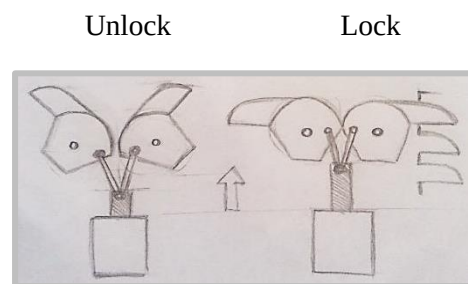
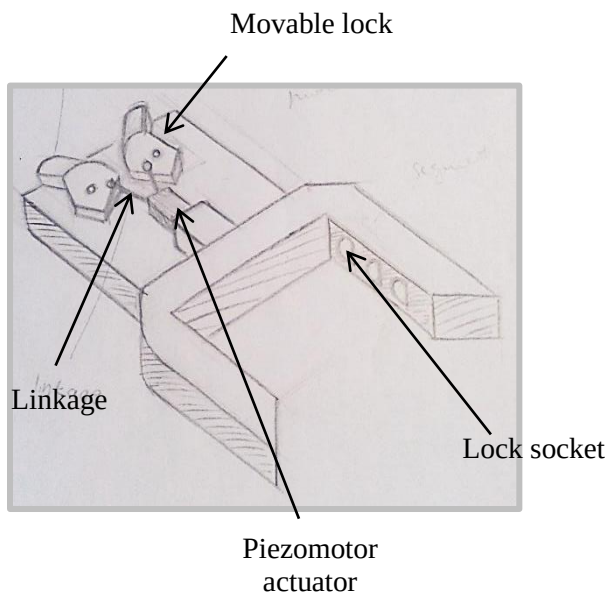
Concept 2



Concept 3



Concept 4



Appendix E Selection Matrix

No.	Selection Criteria	Reference Spinomed	CONCEPTS			
			1	2	3	4
1.0	EASE OF HANDLING (OPERATION)					
	Complexity	0	0	+	+	0
	User-friendliness	0	+	0	+	+
2.0	MAINTENANCE					
	System complexity	0	+	0	0	+
	Durability	0	0	0	0	+
3.0	EASE OF MANUFACTURING					
	3.1 Ready-made	0	+	0	0	+
	3.2 Complexity of part	0	0	0	0	0
4.0	SMOOTHNESS OF OPERATION	0	+	0	+	+
5.0	SAFETY					
	Harness	0	0	0	0	0
	Emergency mechanism	0	+	+	+	+
Sum “+”		0	5	2	4	6
Sum “0”		10	5	8	6	4
Sum “-”		0	0	0	0	0
Net Score		0	5	2	4	6
Rank			2	4	3	1
Conclusion/continue?			Analyze further	Reject	Reject	Continue

Appendix F Rubric for Selection Concept

CONCEPT EVALUATION FORM

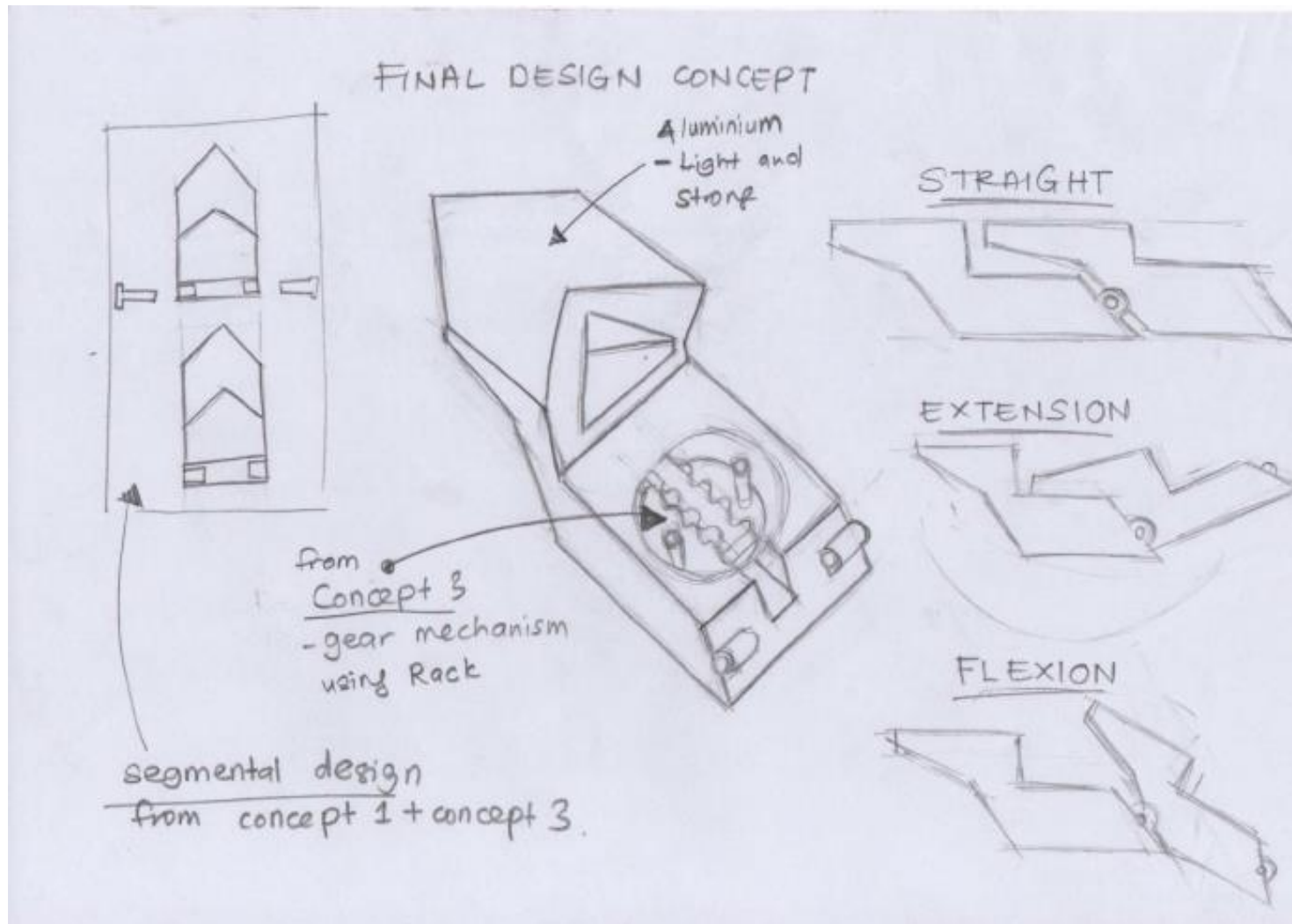
CRITERIA		CONCEPT			
TOTAL = 100%		A	B	C	D
1.0 EASE OF HANDLING (OPERATION) (Total 30%)					
1.1 Operation step: Complexity (15%)	SCALE				
The design and mechanics are basic and uncomplicated.	10~15				
The mechanics and complexity of the design are moderate.	5~9				
The design features a high level of complexity as well as a complicated mechanism.	0~4				
1.2 User-friendliness: Self-operated (15%)	SCALE				
The concept and mechanism are straightforward to be operated.	10~15				
The concept and mechanism are moderately challenging to operate.	5~9				
With such a mechanism and concept, it is tough to operate.	0~4				
2.0 MAINTENANCE (Total 20%)					
2.1 Ease of disassembling/assemble: System complexity (8%)	SCALE				
Components are simple to disassemble and connect to other parts.	6~8				
Components are moderately difficult to dismantle due to their complexity and connectivity.	3~5				
Components having a high level of complexity and connection with other parts are difficult to dismantle.	0~2				
2.2 Durability (12%)					
• Robust material (5%)	SCALE				
This concept has high durability.	4~5				
This concept has a fair amount of durability.	2~3				
In terms of long-term durability, this concept is lacking.	0~1				
• System integrity: stability & strength (7%)	SCALE				
When in use, this concept has a high level of stability and strength, with minimal risk of failure.	6~7				
When in operation, this concept has moderate stability and strength, with a moderate risk of failure.	3~5				
When in use, this concept has a low level of stability and strength, with a significant risk of failure.	0~2				
3.0 EASE OF MANUFACTURING (Total 10%)					
3.1 Ready-made (5%)	SCALE				
The component is readily available and easy to get.	4~5				
The component is only available in a few markets or places	2~3				
The component is difficult or not available in the market, and may need to be fabricated.	0~1				
3.2 Complexity of part (5%)	SCALE				
The concept has a low level of complexity in terms of geometry and design.	4~5				
The concept has a moderate level of complexity in terms of geometry and design.	2~3				
The concept is very complex in geometry and design, thus difficult to fabricate.	0~1				
4.0 SMOOTHNESS OF OPERATION (Total 10%)	SCALE				
This concept has an excellent and smooth mechanism, thus easy to be used.	8~10				
This concept is simple to use and has a reasonably smooth operation.	4~7				
This concept has a difficult operation mechanism.	0~3				

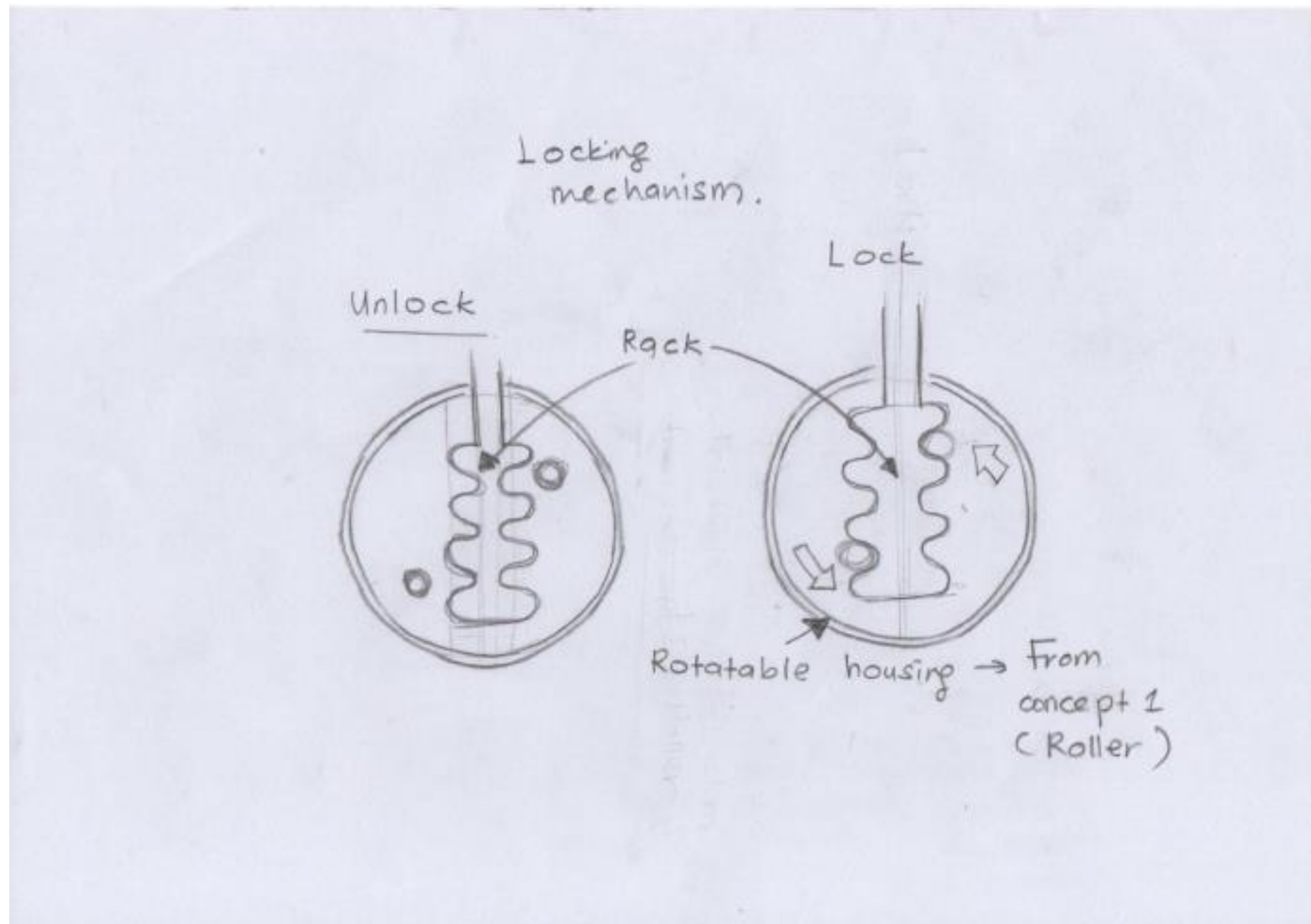
CRITERIA		CONCEPT			
TOTAL = 100%		A	B	C	D
5. SAFETY (20%)					
5.1 Harness (10%)	SCALE				
The harness's design is outstanding, with a great degree of adaptability to the concept design.	8~10				
The harness's design has moderate workability to fit with the concept design.	4~7				
The harness's design is poorly fitted with the concept design.	0~3				
5.2 Emergency mechanism (10%)	SCALE				
The concept has an excellent and reliable emergency mechanism.	8~10				
The concept has a fairly acceptable emergency mechanism.	4~7				
The concept has a poor emergency mechanism.	0~3				
6. FORM/SHAPE (10%)					
6.1 Aesthetic (5%)	SCALE				
The concept has a high aesthetic value and is visually appealing.	4~5				
The concept has a modest aesthetic value and is simple yet appealing to the eye.	2~3				
The concept is unappealing to the eye and has a poor aesthetic aspect.	0~1				
6.2 Simplicity (5%)	SCALE				
The design of the concept is simplistic but functional.	4~5				
The design is not too simple yet functional.	2~3				
The concept's design is very complicated.	0~1				
TOTAL MARK					

Appendix G Selection Concept

		CONCEPT							
		1		2		3		4	
Criteria	Weightage	Rating	W/s	Rating	W/s	Rating	W/s	Rating	W/s
1.0 EASE OF HANDLING (OPERATION) (30%)									
1.1 Operation step									
Complexity	15	10.01	1.50	13.76	2.06	10.01	1.50	7.50	1.13
1.2 User-friendliness									
Self-operated	15	12.99	1.95	9.99	1.50	11.01	1.65	12.99	1.95
2.0 MAINTENANCE (20%)									
2.1 Easy of dissembling/assemble									
System complexity	8	4.80	0.38	4.00	0.32	6.40	0.51	4.27	0.34
2.2 Durability									
Robust material	5	2.67	0.13	4.33	0.22	4.33	0.22	3.33	0.17
System integrity (stability & strength)	7	5.60	0.39	5.14	0.36	7.00	0.49	5.60	0.39
3.0 EASE OF MANUFACTURING (10%)									
3.1 Ready-made	5	4.33	0.22	2.00	0.10	2.67	0.13	4.33	0.22
3.2 Complexity of part	5	3.67	0.18	2.67	0.13	3.67	0.18	3.67	0.18
4.0 SMOOTHNESS OF OPERATION (10%)	10	10.00	1.00	8.00	0.80	9.34	0.93	9.34	0.93
5.0 SAFETY (20%)									
5.1 Harness	10	7.50	0.75	6.68	0.67	8.33	0.83	8.33	0.83
5.2 Emergency mechanism	10	9.18	0.92	9.18	0.92	9.18	0.92	9.18	0.92
6.0 FORM/SHAPE (10%)									
6.1 Aesthetic	5	4.00	0.20	2.33	0.12	3.67	0.18	3.67	0.18
6.2 Simplicity	5	3.67	0.18	3.33	0.17	2.67	0.13	4.00	0.20
Total	100		7.81		7.36		7.69		7.44

Appendix H Sketch for the Final Concept Design





Appendix I Normalize Segmental and Mean Differences during different Forward Bending approaches

Normalize Mean Difference $\pm$ SD Forward Bending: Straight Back (Begin cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	7.24 $\pm$ 3.62	18.52 $\pm$ 7.00	29.90 $\pm$ 6.93	33.43 $\pm$ 4.61	24.38 $\pm$ 13.49	0.76 $\pm$ 16.64	-11.00 $\pm$ 10.58	-10.60 $\pm$ 6.49	-7.60 $\pm$ 4.78	-3.33 $\pm$ 2.58
	T4-6	0.00 $\pm$ 0.00	6.94 $\pm$ 3.74	19.28 $\pm$ 7.33	31.60 $\pm$ 7.10	35.71 $\pm$ 3.92	32.37 $\pm$ 5.49	18.53 $\pm$ 14.39	-2.13 $\pm$ 15.32	-10.38 $\pm$ 6.51	-8.87 $\pm$ 3.85	-3.38 $\pm$ 2.07
	T6-8	0.00 $\pm$ 0.00	3.26 $\pm$ 4.77	15.58 $\pm$ 10.46	30.09 $\pm$ 8.16	35.84 $\pm$ 3.97	33.28 $\pm$ 4.69	25.86 $\pm$ 6.63	12.43 $\pm$ 13.06	-1.54 $\pm$ 11.03	-6.20 $\pm$ 5.42	-3.55 $\pm$ 2.80
	T8-10	0.00 $\pm$ 0.00	-4.85 $\pm$ 4.49	4.01 $\pm$ 13.39	25.59 $\pm$ 15.25	35.18 $\pm$ 4.25	33.56 $\pm$ 4.35	27.29 $\pm$ 6.04	19.68 $\pm$ 7.67	10.57 $\pm$ 9.36	1.51 $\pm$ 8.69	-1.78 $\pm$ 5.66
	Global T1-10	0.00 $\pm$ 0.00	0.54 $\pm$ 1.89	11.71 $\pm$ 9.76	27.39 $\pm$ 10.44	34.54 $\pm$ 3.91	28.05 $\pm$ 10.21	12.46 $\pm$ 11.71	3.12 $\pm$ 5.30	-0.08 $\pm$ 5.22	-4.02 $\pm$ 5.48	-2.10 $\pm$ 2.03
Lumbar	T10-L1	0.00 $\pm$ 0.00	-6.38 $\pm$ 3.45	-6.59 $\pm$ 11.09	18.83 $\pm$ 18.13	34.06 $\pm$ 4.98	32.86 $\pm$ 4.26	27.62 $\pm$ 5.08	21.12 $\pm$ 5.76	12.49 $\pm$ 8.22	6.00 $\pm$ 5.61	1.54 $\pm$ 3.96
	L1-3	0.00 $\pm$ 0.00	-4.11 $\pm$ 4.24	-5.09 $\pm$ 11.70	16.96 $\pm$ 14.02	30.55 $\pm$ 5.28	29.80 $\pm$ 4.21	25.48 $\pm$ 4.92	19.92 $\pm$ 5.82	13.31 $\pm$ 6.04	7.86 $\pm$ 4.44	2.87 $\pm$ 2.68
	L3-5	0.00 $\pm$ 0.00	0.58 $\pm$ 2.95	6.25 $\pm$ 6.21	17.06 $\pm$ 7.28	23.21 $\pm$ 5.49	23.28 $\pm$ 4.73	20.27 $\pm$ 4.29	16.20 $\pm$ 4.69	11.31 $\pm$ 4.33	6.85 $\pm$ 3.34	2.70 $\pm$ 1.99
	Global T10-L5	0.00 $\pm$ 0.00	-2.80 $\pm$ 2.34	-0.59 $\pm$ 7.37	18.65 $\pm$ 11.70	28.64 $\pm$ 5.06	28.16 $\pm$ 3.98	23.96 $\pm$ 4.50	18.78 $\pm$ 5.19	12.42 $\pm$ 5.90	6.15 $\pm$ 4.14	1.67 $\pm$ 2.38

Normalize Mean Difference $\pm$ SD Forward Bending: Straight Back (Return cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	90.81 $\pm$ 23.92	94.53 $\pm$ 22.86	107.49 $\pm$ 22.77	123.42 $\pm$ 22.99	112.40 $\pm$ 18.26	85.96 $\pm$ 18.15	57.41 $\pm$ 18.88	32.27 $\pm$ 12.84	14.57 $\pm$ 7.37	4.16 $\pm$ 3.12	0.00 $\pm$ 0.00
	T4-6	124.13 $\pm$ 19.80	127.82 $\pm$ 19.33	140.26 $\pm$ 17.82	145.99 $\pm$ 14.42	123.65 $\pm$ 17.86	91.23 $\pm$ 20.82	58.84 $\pm$ 19.08	31.76 $\pm$ 14.24	13.22 $\pm$ 8.44	3.51 $\pm$ 3.31	0.00 $\pm$ 0.00
	T6-8	143.76 $\pm$ 14.09	147.50 $\pm$ 13.76	155.17 $\pm$ 10.07	143.90 $\pm$ 16.27	113.23 $\pm$ 22.43	79.36 $\pm$ 26.23	46.82 $\pm$ 24.77	20.50 $\pm$ 20.46	5.15 $\pm$ 12.25	0.66 $\pm$ 3.99	0.00 $\pm$ 0.00
	T8-10	145.70 $\pm$ 13.59	147.20 $\pm$ 12.15	143.74 $\pm$ 14.37	119.93 $\pm$ 19.72	87.37 $\pm$ 24.24	52.58 $\pm$ 27.00	20.16 $\pm$ 26.60	-2.45 $\pm$ 19.12	-8.36 $\pm$ 9.84	-3.16 $\pm$ 3.42	0.00 $\pm$ 0.00
	Global T1-10	114.17 $\pm$ 11.09	116.64 $\pm$ 10.35	121.77 $\pm$ 11.13	118.97 $\pm$ 12.07	98.66 $\pm$ 15.08	68.67 $\pm$ 20.35	38.47 $\pm$ 20.97	14.72 $\pm$ 14.56	2.79 $\pm$ 6.66	0.46 $\pm$ 1.89	0.00 $\pm$ 0.00
Lumbar	T10-L1	138.07 $\pm$ 15.80	137.53 $\pm$ 15.77	127.69 $\pm$ 17.77	102.21 $\pm$ 18.91	70.14 $\pm$ 21.01	36.12 $\pm$ 22.75	4.72 $\pm$ 22.72	-14.15 $\pm$ 14.54	-11.86 $\pm$ 6.76	-3.72 $\pm$ 2.43	0.00 $\pm$ 0.00
	L1-3	135.32 $\pm$ 22.80	131.72 $\pm$ 24.08	118.84 $\pm$ 25.74	95.08 $\pm$ 26.21	66.03 $\pm$ 27.13	35.60 $\pm$ 27.35	7.94 $\pm$ 26.64	-8.15 $\pm$ 20.18	-6.87 $\pm$ 10.05	-2.11 $\pm$ 3.14	0.00 $\pm$ 0.00
	L3-5	127.88 $\pm$ 20.49	123.33 $\pm$ 20.81	110.23 $\pm$ 19.80	89.40 $\pm$ 18.11	65.17 $\pm$ 16.61	41.31 $\pm$ 14.96	21.03 $\pm$ 13.63	6.82 $\pm$ 12.03	1.28 $\pm$ 6.74	0.16 $\pm$ 2.09	0.00 $\pm$ 0.00
	Global T10-L5	133.55 $\pm$ 13.40	131.09 $\pm$ 14.37	119.54 $\pm$ 15.29	96.06 $\pm$ 14.75	67.67 $\pm$ 15.43	38.56 $\pm$ 15.72	12.50 $\pm$ 15.06	-4.44 $\pm$ 10.93	-5.85 $\pm$ 5.34	-1.91 $\pm$ 1.89	0.00 $\pm$ 0.00

Normalize Segmental Difference $\pm$ SD Forward Bending: Straight Back (Begin cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	7.24 $\pm$ 3.62	18.52 $\pm$ 7.00	29.90 $\pm$ 6.93	33.43 $\pm$ 4.61	24.38 $\pm$ 13.49	0.76 $\pm$ 16.64	-11.00 $\pm$ 10.58	-10.60 $\pm$ 6.49	-7.60 $\pm$ 4.78	-3.33 $\pm$ 2.58
	T4-6	0.00 $\pm$ 0.00	6.94 $\pm$ 3.74	19.28 $\pm$ 7.33	31.60 $\pm$ 7.10	35.71 $\pm$ 3.92	32.37 $\pm$ 5.49	18.53 $\pm$ 14.39	-2.13 $\pm$ 15.32	-10.38 $\pm$ 6.51	-8.87 $\pm$ 3.85	-3.38 $\pm$ 2.07
	T6-8	0.00 $\pm$ 0.00	3.26 $\pm$ 4.77	15.58 $\pm$ 10.46	30.09 $\pm$ 8.16	35.84 $\pm$ 3.97	33.28 $\pm$ 4.69	25.86 $\pm$ 6.63	12.43 $\pm$ 13.06	-1.54 $\pm$ 11.03	-6.20 $\pm$ 5.42	-3.55 $\pm$ 2.80
	T8-10	0.00 $\pm$ 0.00	-4.85 $\pm$ 4.49	4.01 $\pm$ 13.39	25.59 $\pm$ 15.25	35.18 $\pm$ 4.25	33.56 $\pm$ 4.35	27.29 $\pm$ 6.04	19.68 $\pm$ 7.67	10.57 $\pm$ 9.36	1.51 $\pm$ 8.69	-1.78 $\pm$ 5.66
	Global T1-10	0.00 $\pm$ 0.00	0.54 $\pm$ 1.89	11.71 $\pm$ 9.76	27.39 $\pm$ 10.44	34.54 $\pm$ 3.91	28.05 $\pm$ 10.21	12.46 $\pm$ 11.71	3.12 $\pm$ 5.30	-0.08 $\pm$ 5.22	-4.02 $\pm$ 5.48	-2.10 $\pm$ 2.03
Lumbar	T10-L1	0.00 $\pm$ 0.00	-6.38 $\pm$ 3.45	-6.59 $\pm$ 11.09	18.83 $\pm$ 18.13	34.06 $\pm$ 4.98	32.86 $\pm$ 4.26	27.62 $\pm$ 5.08	21.12 $\pm$ 5.76	12.49 $\pm$ 8.22	6.00 $\pm$ 5.61	1.54 $\pm$ 3.96
	L1-3	0.00 $\pm$ 0.00	-4.11 $\pm$ 4.24	-5.09 $\pm$ 11.70	16.96 $\pm$ 14.02	30.55 $\pm$ 5.28	29.80 $\pm$ 4.21	25.48 $\pm$ 4.92	19.92 $\pm$ 5.82	13.31 $\pm$ 6.04	7.86 $\pm$ 4.44	2.87 $\pm$ 2.68
	L3-5	0.00 $\pm$ 0.00	0.58 $\pm$ 2.95	6.25 $\pm$ 6.21	17.06 $\pm$ 7.28	23.21 $\pm$ 5.49	23.28 $\pm$ 4.73	20.27 $\pm$ 4.29	16.20 $\pm$ 4.69	11.31 $\pm$ 4.33	6.85 $\pm$ 3.34	2.70 $\pm$ 1.99
	Global T10-L5	0.00 $\pm$ 0.00	-2.80 $\pm$ 2.34	-0.59 $\pm$ 7.37	18.65 $\pm$ 11.70	28.64 $\pm$ 5.06	28.16 $\pm$ 3.98	23.96 $\pm$ 4.50	18.78 $\pm$ 5.19	12.42 $\pm$ 5.90	6.15 $\pm$ 4.14	1.67 $\pm$ 2.38

Normalize Segmental Difference $\pm$ SD Forward Bending: Straight Back (Return cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	3.73 $\pm$ 3.96	12.95 $\pm$ 6.43	15.93 $\pm$ 12.79	-11.02 $\pm$ 19.91	-26.44 $\pm$ 16.09	-28.55 $\pm$ 8.41	-25.15 $\pm$ 6.88	-17.70 $\pm$ 6.23	-10.41 $\pm$ 4.54	-4.16 $\pm$ 3.12
	T4-6	0.00 $\pm$ 0.00	3.69 $\pm$ 3.35	12.44 $\pm$ 7.58	5.73 $\pm$ 20.20	-22.34 $\pm$ 16.70	-32.42 $\pm$ 12.23	-32.39 $\pm$ 4.62	-27.08 $\pm$ 6.50	-18.54 $\pm$ 6.95	-9.71 $\pm$ 5.68	-3.51 $\pm$ 3.31
	T6-8	0.00 $\pm$ 0.00	3.75 $\pm$ 3.28	7.67 $\pm$ 9.76	-11.27 $\pm$ 17.16	-30.68 $\pm$ 10.97	-33.86 $\pm$ 7.30	-32.54 $\pm$ 4.85	-26.32 $\pm$ 6.59	-15.35 $\pm$ 10.41	-4.50 $\pm$ 8.84	-0.66 $\pm$ 3.99
	T8-10	0.00 $\pm$ 0.00	1.51 $\pm$ 4.48	-3.47 $\pm$ 13.26	-23.81 $\pm$ 9.17	-32.57 $\pm$ 7.78	-34.79 $\pm$ 5.20	-32.42 $\pm$ 4.85	-22.62 $\pm$ 12.18	-5.90 $\pm$ 13.38	5.20 $\pm$ 7.75	3.16 $\pm$ 3.42
	Global T1-10	0.00 $\pm$ 0.00	2.47 $\pm$ 3.16	5.13 $\pm$ 7.26	-2.80 $\pm$ 6.76	-20.31 $\pm$ 14.23	-29.99 $\pm$ 10.68	-30.20 $\pm$ 4.65	-23.75 $\pm$ 8.41	-11.93 $\pm$ 9.71	-2.33 $\pm$ 5.14	-0.46 $\pm$ 1.89
Lumbar	T10-L1	0.00 $\pm$ 0.00	-0.53 $\pm$ 5.06	-9.85 $\pm$ 9.11	-25.48 $\pm$ 7.72	-32.07 $\pm$ 6.95	-34.02 $\pm$ 5.25	-31.40 $\pm$ 5.47	-18.87 $\pm$ 14.53	2.29 $\pm$ 11.64	8.14 $\pm$ 5.27	3.72 $\pm$ 2.43
	L1-3	0.00 $\pm$ 0.00	-3.61 $\pm$ 3.84	-12.88 $\pm$ 6.68	-23.76 $\pm$ 7.46	-29.04 $\pm$ 6.67	-30.43 $\pm$ 4.88	-27.66 $\pm$ 5.39	-16.09 $\pm$ 11.94	1.28 $\pm$ 12.75	4.76 $\pm$ 7.25	2.11 $\pm$ 3.14
	L3-5	0.00 $\pm$ 0.00	-4.55 $\pm$ 2.46	-13.10 $\pm$ 4.77	-20.82 $\pm$ 6.26	-24.23 $\pm$ 5.73	-23.86 $\pm$ 4.99	-20.28 $\pm$ 4.52	-14.22 $\pm$ 4.44	-5.54 $\pm$ 6.18	-1.12 $\pm$ 4.83	-0.16 $\pm$ 2.09
	Global T10-L5	0.00 $\pm$ 0.00	-2.47 $\pm$ 2.84	-11.54 $\pm$ 5.70	-23.48 $\pm$ 6.97	-28.39 $\pm$ 6.26	-29.11 $\pm$ 4.70	-26.06 $\pm$ 4.48	-16.94 $\pm$ 8.83	-1.41 $\pm$ 8.08	3.93 $\pm$ 3.80	1.91 $\pm$ 1.89

Normalize Mean Difference $\pm$ SD Forward Bending: Straight Hand (Begin cycle)

	Segment	Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	6.88 $\pm$ 3.48	23.85 $\pm$ 9.51	51.10 $\pm$ 16.72	81.34 $\pm$ 20.81	102.70 $\pm$ 19.94	110.42 $\pm$ 23.71	109.11 $\pm$ 27.47	105.00 $\pm$ 28.28	100.94 $\pm$ 29.07	99.36 $\pm$ 28.95
	T4-6	0.00 $\pm$ 0.00	7.50 $\pm$ 4.52	25.19 $\pm$ 10.64	53.45 $\pm$ 18.00	85.77 $\pm$ 22.76	114.79 $\pm$ 23.91	133.58 $\pm$ 21.72	141.75 $\pm$ 21.13	143.37 $\pm$ 21.58	141.30 $\pm$ 21.28	140.75 $\pm$ 21.67
	T6-8	0.00 $\pm$ 0.00	3.42 $\pm$ 5.68	16.60 $\pm$ 13.29	43.82 $\pm$ 19.63	75.75 $\pm$ 23.17	105.48 $\pm$ 23.97	128.35 $\pm$ 22.38	142.82 $\pm$ 18.90	150.11 $\pm$ 16.12	153.27 $\pm$ 14.17	154.32 $\pm$ 13.33
	T8-10	0.00 $\pm$ 0.00	-3.47 $\pm$ 5.85	-2.50 $\pm$ 17.90	19.31 $\pm$ 26.58	49.79 $\pm$ 29.92	79.27 $\pm$ 30.16	102.86 $\pm$ 28.48	119.88 $\pm$ 26.26	131.67 $\pm$ 25.19	138.89 $\pm$ 23.85	141.45 $\pm$ 22.05
	Global T1-10	0.00 $\pm$ 0.00	1.90 $\pm$ 5.48	10.63 $\pm$ 12.97	35.28 $\pm$ 20.56	65.30 $\pm$ 23.47	90.13 $\pm$ 20.99	105.67 $\pm$ 16.89	111.29 $\pm$ 13.99	113.90 $\pm$ 13.05	115.48 $\pm$ 12.88	115.48 $\pm$ 13.60
Lumbar	T10-L1	0.00 $\pm$ 0.00	-5.00 $\pm$ 4.36	-12.25 $\pm$ 14.54	1.95 $\pm$ 23.69	31.25 $\pm$ 26.20	59.73 $\pm$ 25.99	83.19 $\pm$ 24.81	100.70 $\pm$ 23.16	112.48 $\pm$ 21.82	120.73 $\pm$ 22.10	125.08 $\pm$ 22.31
	L1-3	0.00 $\pm$ 0.00	-3.88 $\pm$ 3.95	-8.51 $\pm$ 14.54	2.04 $\pm$ 23.56	27.72 $\pm$ 25.06	53.84 $\pm$ 24.50	75.72 $\pm$ 23.18	92.63 $\pm$ 22.70	104.81 $\pm$ 23.14	112.78 $\pm$ 24.27	116.80 $\pm$ 25.35
	L3-5	0.00 $\pm$ 0.00	0.17 $\pm$ 3.21	4.39 $\pm$ 9.78	18.69 $\pm$ 15.64	38.12 $\pm$ 19.41	58.06 $\pm$ 21.03	75.42 $\pm$ 20.95	89.22 $\pm$ 20.75	99.23 $\pm$ 21.22	106.02 $\pm$ 22.26	109.47 $\pm$ 22.85
	Global T10-L5	0.00 $\pm$ 0.00	-2.03 $\pm$ 3.30	-4.17 $\pm$ 9.96	10.01 $\pm$ 17.93	34.40 $\pm$ 21.34	58.76 $\pm$ 21.61	79.25 $\pm$ 20.31	95.15 $\pm$ 19.34	106.32 $\pm$ 18.91	113.85 $\pm$ 19.40	117.75 $\pm$ 19.84

Normalize Mean Difference $\pm$ SD Forward Bending: Straight Hand (Return cycle)

	Segment	Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	105.40 $\pm$ 30.19	107.68 $\pm$ 28.46	115.73 $\pm$ 24.98	123.67 $\pm$ 18.24	109.62 $\pm$ 16.02	79.44 $\pm$ 13.87	50.29 $\pm$ 10.53	26.74 $\pm$ 7.10	11.40 $\pm$ 4.04	3.18 $\pm$ 1.82	0.00 $\pm$ 0.00
	T4-6	144.39 $\pm$ 18.78	145.75 $\pm$ 17.80	150.07 $\pm$ 15.61	141.54 $\pm$ 17.76	113.06 $\pm$ 17.48	80.81 $\pm$ 15.79	50.78 $\pm$ 13.12	26.32 $\pm$ 9.48	10.21 $\pm$ 6.21	2.43 $\pm$ 1.71	0.00 $\pm$ 0.00
	T6-8	153.50 $\pm$ 13.48	153.31 $\pm$ 14.38	149.46 $\pm$ 18.58	129.44 $\pm$ 20.37	99.94 $\pm$ 20.81	67.42 $\pm$ 20.41	37.53 $\pm$ 19.20	13.62 $\pm$ 17.19	2.27 $\pm$ 10.75	-0.02 $\pm$ 3.30	0.00 $\pm$ 0.00
	T8-10	138.71 $\pm$ 23.60	136.81 $\pm$ 25.08	125.64 $\pm$ 25.43	102.67 $\pm$ 24.64	72.86 $\pm$ 24.82	40.61 $\pm$ 24.49	11.08 $\pm$ 23.64	-7.78 $\pm$ 16.58	-8.29 $\pm$ 6.70	-2.79 $\pm$ 2.05	0.00 $\pm$ 0.00
	Global T1-10	116.92 $\pm$ 13.72	117.02 $\pm$ 13.46	116.34 $\pm$ 12.85	110.80 $\pm$ 13.26	91.17 $\pm$ 15.29	59.75 $\pm$ 14.48	30.57 $\pm$ 13.19	9.34 $\pm$ 8.58	1.34 $\pm$ 3.28	0.21 $\pm$ 1.14	0.00 $\pm$ 0.00
Lumbar	T10-L1	123.58 $\pm$ 25.20	120.01 $\pm$ 25.60	107.64 $\pm$ 26.12	84.77 $\pm$ 26.40	55.63 $\pm$ 26.11	24.64 $\pm$ 24.83	-2.59 $\pm$ 22.00	-15.84 $\pm$ 12.53	-10.40 $\pm$ 5.81	-3.01 $\pm$ 1.84	0.00 $\pm$ 0.00
	L1-3	116.92 $\pm$ 28.39	112.86 $\pm$ 28.79	100.33 $\pm$ 28.77	78.59 $\pm$ 28.75	51.64 $\pm$ 28.36	23.53 $\pm$ 27.50	-0.02 $\pm$ 24.95	-9.60 $\pm$ 16.55	-6.88 $\pm$ 6.99	-2.92 $\pm$ 2.05	0.00 $\pm$ 0.00
	L3-5	112.01 $\pm$ 23.45	107.94 $\pm$ 23.81	96.88 $\pm$ 23.54	78.37 $\pm$ 22.31	56.19 $\pm$ 19.99	34.21 $\pm$ 17.06	16.34 $\pm$ 13.86	5.14 $\pm$ 9.30	0.43 $\pm$ 5.49	-0.14 $\pm$ 1.89	0.00 $\pm$ 0.00
	Global T10-L5	118.29 $\pm$ 21.23	114.29 $\pm$ 21.71	102.41 $\pm$ 21.86	81.46 $\pm$ 21.32	55.60 $\pm$ 19.90	28.96 $\pm$ 17.69	5.96 $\pm$ 15.11	-5.89 $\pm$ 8.91	-4.81 $\pm$ 4.36	-1.54 $\pm$ 1.48	0.00 $\pm$ 0.00

Normalize Segmental Difference $\pm$ SD Forward Bending: Straight Hand (Begin cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	6.88 $\pm$ 3.48	16.97 $\pm$ 6.46	27.24 $\pm$ 8.18	30.24 $\pm$ 7.23	21.36 $\pm$ 12.77	7.72 $\pm$ 16.93	-1.31 $\pm$ 11.73	-4.11 $\pm$ 8.05	-4.06 $\pm$ 3.90	-1.59 $\pm$ 2.85
	T4-6	0.00 $\pm$ 0.00	7.50 $\pm$ 4.52	17.69 $\pm$ 7.49	28.25 $\pm$ 8.61	32.32 $\pm$ 7.06	29.02 $\pm$ 6.59	18.79 $\pm$ 10.25	8.17 $\pm$ 12.27	1.62 $\pm$ 9.22	-2.07 $\pm$ 5.77	-0.55 $\pm$ 2.92
	T6-8	0.00 $\pm$ 0.00	3.42 $\pm$ 5.68	13.17 $\pm$ 9.95	27.22 $\pm$ 9.06	31.94 $\pm$ 6.79	29.72 $\pm$ 5.31	22.87 $\pm$ 5.55	14.47 $\pm$ 7.35	7.30 $\pm$ 7.73	3.16 $\pm$ 5.97	1.05 $\pm$ 3.06
	T8-10	0.00 $\pm$ 0.00	-3.47 $\pm$ 5.85	0.97 $\pm$ 13.87	21.81 $\pm$ 13.77	30.48 $\pm$ 7.80	29.48 $\pm$ 5.53	23.59 $\pm$ 5.11	17.01 $\pm$ 5.48	11.79 $\pm$ 4.73	7.22 $\pm$ 3.46	2.56 $\pm$ 2.94
	Global T1-10	0.00 $\pm$ 0.00	1.90 $\pm$ 5.48	8.73 $\pm$ 9.32	24.65 $\pm$ 10.23	30.02 $\pm$ 7.26	24.83 $\pm$ 8.64	15.54 $\pm$ 10.69	5.62 $\pm$ 7.73	2.61 $\pm$ 4.57	1.58 $\pm$ 3.13	-0.01 $\pm$ 1.69
Lumbar	T10-L1	0.00 $\pm$ 0.00	-5.00 $\pm$ 4.36	-7.25 $\pm$ 11.79	14.20 $\pm$ 16.85	29.31 $\pm$ 7.16	28.48 $\pm$ 4.86	23.46 $\pm$ 4.35	17.51 $\pm$ 4.80	11.78 $\pm$ 4.67	8.25 $\pm$ 4.13	4.35 $\pm$ 2.86
	L1-3	0.00 $\pm$ 0.00	-3.88 $\pm$ 3.95	-4.63 $\pm$ 11.69	10.55 $\pm$ 16.07	25.69 $\pm$ 7.00	26.12 $\pm$ 4.80	21.87 $\pm$ 4.99	16.91 $\pm$ 4.60	12.18 $\pm$ 4.02	7.97 $\pm$ 3.44	4.01 $\pm$ 2.30
	L3-5	0.00 $\pm$ 0.00	0.17 $\pm$ 3.21	4.22 $\pm$ 7.13	14.31 $\pm$ 7.63	19.42 $\pm$ 5.81	19.94 $\pm$ 4.40	17.36 $\pm$ 4.20	13.80 $\pm$ 4.24	10.01 $\pm$ 3.64	6.79 $\pm$ 2.56	3.45 $\pm$ 1.41
	Global T10-L5	0.00 $\pm$ 0.00	-2.03 $\pm$ 3.30	-2.14 $\pm$ 7.65	14.18 $\pm$ 11.70	24.38 $\pm$ 6.27	24.37 $\pm$ 4.17	20.48 $\pm$ 4.27	15.91 $\pm$ 4.05	11.16 $\pm$ 3.80	7.53 $\pm$ 3.01	3.90 $\pm$ 1.66

Normalize Segmental Difference $\pm$ SD Forward Bending: Straight Hand (Return cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	2.28 $\pm$ 3.06	8.05 $\pm$ 5.82	7.94 $\pm$ 13.67	-14.06 $\pm$ 15.41	-30.18 $\pm$ 5.66	-29.14 $\pm$ 4.71	-23.55 $\pm$ 4.06	-15.34 $\pm$ 3.44	-8.22 $\pm$ 2.47	-3.18 $\pm$ 1.82
	T4-6	0.00 $\pm$ 0.00	1.36 $\pm$ 2.87	4.32 $\pm$ 10.03	-8.52 $\pm$ 13.92	-28.48 $\pm$ 6.31	-32.25 $\pm$ 5.01	-30.04 $\pm$ 4.34	-24.46 $\pm$ 4.51	-16.11 $\pm$ 4.66	-7.78 $\pm$ 5.20	-2.43 $\pm$ 1.71
	T6-8	0.00 $\pm$ 0.00	-0.19 $\pm$ 3.99	-3.84 $\pm$ 9.39	-20.03 $\pm$ 6.90	-29.50 $\pm$ 5.36	-32.52 $\pm$ 5.05	-29.89 $\pm$ 4.66	-23.90 $\pm$ 4.86	-11.35 $\pm$ 8.26	-2.29 $\pm$ 7.61	0.02 $\pm$ 3.30
	T8-10	0.00 $\pm$ 0.00	-1.90 $\pm$ 3.08	-11.17 $\pm$ 4.96	-22.96 $\pm$ 5.76	-29.81 $\pm$ 5.35	-32.25 $\pm$ 4.95	-29.52 $\pm$ 4.64	-18.87 $\pm$ 10.37	-0.51 $\pm$ 12.62	5.50 $\pm$ 5.11	2.79 $\pm$ 2.05
	Global T1-10	0.00 $\pm$ 0.00	0.11 $\pm$ 2.11	-0.68 $\pm$ 2.53	-5.55 $\pm$ 7.66	-19.63 $\pm$ 10.99	-31.42 $\pm$ 5.30	-29.19 $\pm$ 4.16	-21.22 $\pm$ 6.16	-8.01 $\pm$ 7.18	-1.13 $\pm$ 2.42	-0.21 $\pm$ 1.14
Lumbar	T10-L1	0.00 $\pm$ 0.00	-3.57 $\pm$ 2.89	-12.38 $\pm$ 4.97	-22.87 $\pm$ 5.38	-29.14 $\pm$ 5.42	-30.99 $\pm$ 4.92	-27.23 $\pm$ 6.51	-13.25 $\pm$ 14.17	5.44 $\pm$ 9.42	7.39 $\pm$ 4.33	3.01 $\pm$ 1.84
	L1-3	0.00 $\pm$ 0.00	-4.07 $\pm$ 2.90	-12.53 $\pm$ 4.65	-21.74 $\pm$ 5.48	-26.95 $\pm$ 5.22	-28.10 $\pm$ 4.99	-23.56 $\pm$ 6.74	-9.58 $\pm$ 13.54	2.72 $\pm$ 10.77	3.96 $\pm$ 5.52	2.92 $\pm$ 2.05
	L3-5	0.00 $\pm$ 0.00	-4.07 $\pm$ 2.08	-11.06 $\pm$ 3.58	-18.51 $\pm$ 4.73	-22.19 $\pm$ 5.31	-21.97 $\pm$ 5.42	-17.87 $\pm$ 4.86	-11.20 $\pm$ 6.36	-4.70 $\pm$ 4.81	-0.58 $\pm$ 3.89	0.14 $\pm$ 1.89
	Global T10-L5	0.00 $\pm$ 0.00	-4.00 $\pm$ 2.38	-11.89 $\pm$ 4.14	-20.94 $\pm$ 5.03	-25.86 $\pm$ 5.07	-26.65 $\pm$ 4.92	-22.99 $\pm$ 4.76	-11.85 $\pm$ 9.55	1.07 $\pm$ 6.57	3.28 $\pm$ 3.09	1.54 $\pm$ 1.48

Normalize Mean Difference $\pm$ SD for Forward Bending: Hanging Hand (Begin cycle)

	Segment	Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	6.63 $\pm$ 3.08	23.27 $\pm$ 8.54	50.75 $\pm$ 14.88	83.02 $\pm$ 15.86	107.52 $\pm$ 12.99	108.71 $\pm$ 23.49	91.17 $\pm$ 25.25	75.01 $\pm$ 24.43	63.98 $\pm$ 24.45	58.88 $\pm$ 24.93
	T4-6	0.00 $\pm$ 0.00	6.37 $\pm$ 3.03	23.23 $\pm$ 8.67	51.37 $\pm$ 14.72	85.97 $\pm$ 19.10	118.28 $\pm$ 16.41	138.87 $\pm$ 13.84	133.99 $\pm$ 21.70	117.89 $\pm$ 21.66	107.57 $\pm$ 21.76	103.33 $\pm$ 21.67
	T6-8	0.00 $\pm$ 0.00	3.32 $\pm$ 4.84	15.43 $\pm$ 13.09	42.28 $\pm$ 17.34	76.53 $\pm$ 20.49	111.58 $\pm$ 20.39	139.77 $\pm$ 13.90	155.05 $\pm$ 11.69	149.75 $\pm$ 15.28	139.66 $\pm$ 16.38	136.36 $\pm$ 17.08
	T8-10	0.00 $\pm$ 0.00	-5.10 $\pm$ 2.98	-4.64 $\pm$ 12.77	16.58 $\pm$ 22.36	50.02 $\pm$ 23.77	84.96 $\pm$ 23.28	115.94 $\pm$ 19.63	138.51 $\pm$ 13.92	149.58 $\pm$ 13.48	148.81 $\pm$ 14.06	147.49 $\pm$ 15.34
	Global T1-10	0.00 $\pm$ 0.00	0.35 $\pm$ 1.13	9.14 $\pm$ 8.93	33.48 $\pm$ 16.37	66.10 $\pm$ 16.96	95.99 $\pm$ 11.50	109.87 $\pm$ 10.68	110.94 $\pm$ 14.32	108.85 $\pm$ 16.88	103.28 $\pm$ 17.62	100.01 $\pm$ 18.23
Lumbar	T10-L1	0.00 $\pm$ 0.00	-5.50 $\pm$ 2.59	-14.41 $\pm$ 9.54	-1.49 $\pm$ 19.27	29.58 $\pm$ 21.52	62.78 $\pm$ 21.54	93.31 $\pm$ 20.15	117.65 $\pm$ 16.46	133.70 $\pm$ 14.14	140.09 $\pm$ 13.30	140.41 $\pm$ 12.09
	L1-3	0.00 $\pm$ 0.00	-3.50 $\pm$ 4.13	-9.40 $\pm$ 11.78	-1.18 $\pm$ 19.49	25.62 $\pm$ 18.50	54.58 $\pm$ 17.72	82.03 $\pm$ 17.55	105.40 $\pm$ 18.57	122.13 $\pm$ 20.58	131.77 $\pm$ 22.22	134.84 $\pm$ 22.27
	L3-5	0.00 $\pm$ 0.00	0.75 $\pm$ 3.16	4.01 $\pm$ 10.00	17.69 $\pm$ 13.03	37.19 $\pm$ 14.17	58.21 $\pm$ 15.26	79.16 $\pm$ 16.50	98.55 $\pm$ 17.43	113.36 $\pm$ 18.36	121.89 $\pm$ 19.07	124.91 $\pm$ 19.32
	Global T10-L5	0.00 $\pm$ 0.00	-2.61 $\pm$ 1.73	-5.45 $\pm$ 7.33	8.34 $\pm$ 12.76	33.82 $\pm$ 15.07	61.13 $\pm$ 15.96	86.98 $\pm$ 15.99	108.78 $\pm$ 14.84	124.66 $\pm$ 14.76	132.46 $\pm$ 13.63	134.29 $\pm$ 12.83

Normalize Mean Difference $\pm$ SD for Forward Bending: Hanging Hand (Return cycle)

	Segment	Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	60.66 $\pm$ 27.77	63.96 $\pm$ 27.48	77.13 $\pm$ 28.41	100.44 $\pm$ 26.24	114.15 $\pm$ 19.69	93.82 $\pm$ 18.95	63.08 $\pm$ 18.77	35.45 $\pm$ 13.34	16.21 $\pm$ 7.76	4.95 $\pm$ 3.40	0.00 $\pm$ 0.00
	T4-6	102.59 $\pm$ 23.79	106.11 $\pm$ 23.59	119.62 $\pm$ 23.02	138.72 $\pm$ 16.76	132.08 $\pm$ 17.22	97.27 $\pm$ 19.26	63.27 $\pm$ 16.90	35.15 $\pm$ 12.67	15.40 $\pm$ 7.49	4.34 $\pm$ 3.18	0.00 $\pm$ 0.00
	T6-8	133.37 $\pm$ 15.92	137.24 $\pm$ 15.86	149.65 $\pm$ 12.51	152.70 $\pm$ 9.82	124.52 $\pm$ 18.27	87.53 $\pm$ 18.97	53.56 $\pm$ 17.84	26.13 $\pm$ 14.72	9.13 $\pm$ 9.73	2.21 $\pm$ 3.40	0.00 $\pm$ 0.00
	T8-10	142.20 $\pm$ 15.99	144.70 $\pm$ 14.53	149.01 $\pm$ 11.29	129.02 $\pm$ 17.83	93.73 $\pm$ 22.46	56.72 $\pm$ 25.23	23.76 $\pm$ 25.38	0.96 $\pm$ 19.12	-6.38 $\pm$ 9.15	-3.17 $\pm$ 3.07	0.00 $\pm$ 0.00
	Global T1-10	98.54 $\pm$ 17.79	101.30 $\pm$ 17.38	109.71 $\pm$ 14.24	111.36 $\pm$ 11.31	101.70 $\pm$ 13.44	75.08 $\pm$ 19.00	42.98 $\pm$ 19.34	17.68 $\pm$ 13.81	4.57 $\pm$ 6.02	0.60 $\pm$ 1.30	0.00 $\pm$ 0.00
Lumbar	T10-L1	139.81 $\pm$ 12.20	139.99 $\pm$ 14.25	131.38 $\pm$ 18.29	104.72 $\pm$ 20.26	70.60 $\pm$ 22.36	35.52 $\pm$ 23.46	4.84 $\pm$ 22.76	-12.63 $\pm$ 14.84	-11.25 $\pm$ 6.39	-3.96 $\pm$ 2.29	0.00 $\pm$ 0.00
	L1-3	135.25 $\pm$ 23.26	131.21 $\pm$ 24.01	116.72 $\pm$ 24.76	91.24 $\pm$ 25.90	60.55 $\pm$ 26.66	29.68 $\pm$ 26.29	3.42 $\pm$ 24.87	-10.16 $\pm$ 18.07	-8.59 $\pm$ 7.46	-3.48 $\pm$ 2.41	0.00 $\pm$ 0.00
	L3-5	127.07 $\pm$ 21.00	122.47 $\pm$ 21.79	108.47 $\pm$ 22.17	85.91 $\pm$ 21.88	60.62 $\pm$ 19.59	36.93 $\pm$ 16.83	17.92 $\pm$ 14.51	5.46 $\pm$ 11.06	0.82 $\pm$ 6.13	-0.06 $\pm$ 2.29	0.00 $\pm$ 0.00
	Global T10-L5	134.06 $\pm$ 13.34	132.17 $\pm$ 14.73	121.58 $\pm$ 17.68	96.34 $\pm$ 18.42	66.37 $\pm$ 18.64	36.71 $\pm$ 17.87	11.55 $\pm$ 16.12	-3.95 $\pm$ 10.44	-5.55 $\pm$ 4.58	-1.98 $\pm$ 1.60	0.00 $\pm$ 0.00

Normalize Segmental Difference $\pm$ SD for Forward Bending: Hanging Hand (Begin cycle)

	Segment	Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	6.63 $\pm$ 3.08	16.64 $\pm$ 5.88	27.48 $\pm$ 7.94	32.27 $\pm$ 8.82	24.50 $\pm$ 16.86	1.19 $\pm$ 20.85	-17.54 $\pm$ 7.49	-16.16 $\pm$ 5.20	-11.04 $\pm$ 3.71	-5.10 $\pm$ 3.00
	T4-6	0.00 $\pm$ 0.00	6.37 $\pm$ 3.03	16.86 $\pm$ 6.02	28.14 $\pm$ 7.77	34.61 $\pm$ 7.30	32.30 $\pm$ 11.14	20.59 $\pm$ 21.31	-4.87 $\pm$ 18.22	-16.11 $\pm$ 5.99	-10.32 $\pm$ 3.84	-4.24 $\pm$ 2.35
	T6-8	0.00 $\pm$ 0.00	3.32 $\pm$ 4.84	12.11 $\pm$ 8.67	26.85 $\pm$ 7.23	34.25 $\pm$ 7.64	35.05 $\pm$ 5.75	28.19 $\pm$ 13.13	15.28 $\pm$ 15.53	-5.31 $\pm$ 12.28	-10.09 $\pm$ 5.49	-3.30 $\pm$ 3.11
	T8-10	0.00 $\pm$ 0.00	-5.10 $\pm$ 2.98	0.47 $\pm$ 12.10	21.22 $\pm$ 12.59	33.44 $\pm$ 7.09	34.94 $\pm$ 5.32	30.98 $\pm$ 6.38	22.57 $\pm$ 13.81	11.07 $\pm$ 10.79	-0.77 $\pm$ 7.96	-1.32 $\pm$ 3.53
	Global T1-10	0.00 $\pm$ 0.00	0.35 $\pm$ 1.13	8.79 $\pm$ 8.48	24.34 $\pm$ 8.82	32.61 $\pm$ 7.25	29.89 $\pm$ 9.81	13.88 $\pm$ 12.82	1.07 $\pm$ 6.86	-2.09 $\pm$ 4.80	-5.57 $\pm$ 4.77	-3.27 $\pm$ 2.59
Lumbar	T10-L1	0.00 $\pm$ 0.00	-5.50 $\pm$ 2.59	-8.91 $\pm$ 8.93	12.91 $\pm$ 14.57	31.07 $\pm$ 6.59	33.20 $\pm$ 4.81	30.54 $\pm$ 5.26	24.34 $\pm$ 7.08	16.05 $\pm$ 8.17	6.39 $\pm$ 6.20	0.32 $\pm$ 3.08
	L1-3	0.00 $\pm$ 0.00	-3.50 $\pm$ 4.13	-5.90 $\pm$ 8.50	8.23 $\pm$ 12.25	26.80 $\pm$ 5.52	28.96 $\pm$ 4.58	27.45 $\pm$ 5.02	23.37 $\pm$ 6.07	16.73 $\pm$ 5.74	9.64 $\pm$ 4.28	3.07 $\pm$ 2.66
	L3-5	0.00 $\pm$ 0.00	0.75 $\pm$ 3.16	3.27 $\pm$ 7.13	13.67 $\pm$ 5.14	19.50 $\pm$ 4.89	21.03 $\pm$ 5.01	20.94 $\pm$ 4.99	19.39 $\pm$ 5.37	14.81 $\pm$ 5.24	8.53 $\pm$ 3.64	3.02 $\pm$ 1.85
	Global T10-L5	0.00 $\pm$ 0.00	-2.61 $\pm$ 1.73	-2.85 $\pm$ 6.85	13.79 $\pm$ 9.17	25.48 $\pm$ 5.15	27.31 $\pm$ 4.45	25.85 $\pm$ 4.79	21.80 $\pm$ 6.33	15.88 $\pm$ 6.80	7.80 $\pm$ 5.05	1.82 $\pm$ 2.27

Normalize Segmental Difference $\pm$ SD for Forward Bending: Hanging Hand (Return cycle)

	Segment	Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	3.30 $\pm$ 3.44	13.17 $\pm$ 5.83	23.31 $\pm$ 10.25	13.72 $\pm$ 24.75	-20.34 $\pm$ 21.15	-30.74 $\pm$ 8.41	-27.63 $\pm$ 6.50	-19.24 $\pm$ 6.35	-11.26 $\pm$ 4.65	-4.95 $\pm$ 3.40
	T4-6	0.00 $\pm$ 0.00	3.53 $\pm$ 2.21	13.51 $\pm$ 5.79	19.10 $\pm$ 14.51	-6.64 $\pm$ 26.69	-34.80 $\pm$ 9.12	-34.01 $\pm$ 5.06	-28.12 $\pm$ 5.52	-19.75 $\pm$ 5.99	-11.05 $\pm$ 4.64	-4.34 $\pm$ 3.18
	T6-8	0.00 $\pm$ 0.00	3.86 $\pm$ 2.30	12.41 $\pm$ 8.37	3.05 $\pm$ 17.97	-28.19 $\pm$ 13.46	-36.98 $\pm$ 5.20	-33.98 $\pm$ 5.05	-27.43 $\pm$ 5.98	-16.99 $\pm$ 7.30	-6.92 $\pm$ 6.92	-2.21 $\pm$ 3.40
	T8-10	0.00 $\pm$ 0.00	2.51 $\pm$ 4.12	4.31 $\pm$ 12.14	-20.00 $\pm$ 12.46	-35.28 $\pm$ 7.93	-37.01 $\pm$ 5.48	-32.96 $\pm$ 4.77	-22.80 $\pm$ 11.79	-7.34 $\pm$ 13.90	3.21 $\pm$ 7.85	3.17 $\pm$ 3.07
	Global T1-10	0.00 $\pm$ 0.00	2.76 $\pm$ 2.82	8.41 $\pm$ 8.28	1.64 $\pm$ 8.25	-9.66 $\pm$ 13.79	-26.62 $\pm$ 13.25	-32.10 $\pm$ 5.64	-25.29 $\pm$ 8.28	-13.11 $\pm$ 9.07	-3.97 $\pm$ 5.19	-0.60 $\pm$ 1.30
Lumbar	T10-L1	0.00 $\pm$ 0.00	0.18 $\pm$ 4.91	-8.61 $\pm$ 8.00	-26.66 $\pm$ 7.90	-34.11 $\pm$ 6.26	-35.08 $\pm$ 4.96	-30.68 $\pm$ 5.25	-17.47 $\pm$ 14.00	1.38 $\pm$ 12.35	7.29 $\pm$ 5.20	3.96 $\pm$ 2.29
	L1-3	0.00 $\pm$ 0.00	-4.04 $\pm$ 2.63	-14.48 $\pm$ 4.96	-25.49 $\pm$ 6.44	-30.69 $\pm$ 5.48	-30.87 $\pm$ 4.65	-26.26 $\pm$ 5.57	-13.58 $\pm$ 11.22	1.56 $\pm$ 12.29	5.12 $\pm$ 5.85	3.48 $\pm$ 2.41
	L3-5	0.00 $\pm$ 0.00	-4.60 $\pm$ 2.94	-14.00 $\pm$ 5.09	-22.56 $\pm$ 5.24	-25.28 $\pm$ 5.11	-23.70 $\pm$ 5.36	-19.01 $\pm$ 4.94	-12.46 $\pm$ 5.87	-4.64 $\pm$ 6.45	-0.88 $\pm$ 4.22	0.06 $\pm$ 2.29
	Global T10-L5	0.00 $\pm$ 0.00	-1.89 $\pm$ 2.37	-10.59 $\pm$ 6.11	-25.24 $\pm$ 6.46	-29.97 $\pm$ 5.33	-29.66 $\pm$ 4.62	-25.17 $\pm$ 4.46	-15.50 $\pm$ 9.57	-1.60 $\pm$ 8.51	3.57 $\pm$ 3.53	1.98 $\pm$ 1.60

Appendix J Normalize Segmental and Mean Differences during different Pick object approaches

Normalize Mean Difference $\pm$ SD for Pick object: Stoop (Begin cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	8.68 $\pm$ 8.62	25.66 $\pm$ 16.78	51.52 $\pm$ 24.00	80.14 $\pm$ 28.71	99.76 $\pm$ 35.63	90.92 $\pm$ 35.50	73.78 $\pm$ 28.58	62.29 $\pm$ 25.80	54.45 $\pm$ 28.69	51.23 $\pm$ 28.71
	T4-6	0.00 $\pm$ 0.00	8.17 $\pm$ 9.34	26.99 $\pm$ 20.99	55.38 $\pm$ 28.80	87.56 $\pm$ 31.56	115.51 $\pm$ 36.97	117.04 $\pm$ 39.80	98.11 $\pm$ 32.95	82.84 $\pm$ 28.16	72.16 $\pm$ 28.66	68.81 $\pm$ 29.78
	T6-8	0.00 $\pm$ 0.00	5.23 $\pm$ 10.43	20.94 $\pm$ 24.62	50.14 $\pm$ 35.25	83.71 $\pm$ 34.33	116.44 $\pm$ 36.00	130.91 $\pm$ 34.37	123.52 $\pm$ 31.86	107.20 $\pm$ 26.82	95.12 $\pm$ 26.70	91.37 $\pm$ 28.03
	T8-10	0.00 $\pm$ 0.00	-1.09 $\pm$ 10.95	3.70 $\pm$ 28.25	27.57 $\pm$ 43.19	62.19 $\pm$ 44.38	96.92 $\pm$ 39.76	122.98 $\pm$ 34.34	131.13 $\pm$ 28.51	119.93 $\pm$ 26.31	108.88 $\pm$ 29.49	105.36 $\pm$ 28.43
	Global T1-10	0.00 $\pm$ 0.00	3.66 $\pm$ 9.19	14.30 $\pm$ 20.48	39.39 $\pm$ 30.19	70.55 $\pm$ 30.32	97.16 $\pm$ 27.85	105.14 $\pm$ 27.22	100.81 $\pm$ 25.52	89.45 $\pm$ 22.20	80.68 $\pm$ 25.00	77.44 $\pm$ 24.99
Lumbar	T10-L1	0.00 $\pm$ 0.00	-4.85 $\pm$ 10.39	-6.51 $\pm$ 22.94	11.32 $\pm$ 37.33	44.86 $\pm$ 42.83	80.28 $\pm$ 39.14	109.72 $\pm$ 35.05	127.35 $\pm$ 30.35	123.72 $\pm$ 28.01	115.30 $\pm$ 32.81	112.89 $\pm$ 29.75
	L1-3	0.00 $\pm$ 0.00	-3.07 $\pm$ 9.91	-2.58 $\pm$ 21.06	12.07 $\pm$ 34.35	43.25 $\pm$ 38.72	75.63 $\pm$ 35.89	105.33 $\pm$ 34.15	128.87 $\pm$ 32.66	133.53 $\pm$ 30.38	127.96 $\pm$ 33.80	125.20 $\pm$ 36.10
	L3-5	0.00 $\pm$ 0.00	2.15 $\pm$ 7.47	9.22 $\pm$ 17.19	25.44 $\pm$ 27.21	48.66 $\pm$ 34.06	74.61 $\pm$ 36.29	100.37 $\pm$ 35.15	122.11 $\pm$ 30.37	134.60 $\pm$ 27.99	137.94 $\pm$ 30.00	136.10 $\pm$ 33.20
	Global T10-L5	0.00 $\pm$ 0.00	-1.17 $\pm$ 8.22	1.30 $\pm$ 19.83	18.45 $\pm$ 31.94	47.28 $\pm$ 37.49	78.09 $\pm$ 36.28	105.86 $\pm$ 32.44	125.23 $\pm$ 26.60	129.89 $\pm$ 24.53	127.04 $\pm$ 28.28	125.56 $\pm$ 28.82

Normalize Mean Difference $\pm$ SD for Pick object: Stoop (Return cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	53.29 $\pm$ 20.40	56.09 $\pm$ 20.18	67.62 $\pm$ 20.29	89.54 $\pm$ 22.23	114.58 $\pm$ 18.59	109.81 $\pm$ 16.94	75.64 $\pm$ 20.40	41.93 $\pm$ 14.23	18.91 $\pm$ 7.57	5.49 $\pm$ 2.79	0.00 $\pm$ 0.00
	T4-6	68.00 $\pm$ 17.23	71.51 $\pm$ 17.85	85.30 $\pm$ 21.61	111.43 $\pm$ 27.42	134.77 $\pm$ 19.20	122.30 $\pm$ 23.65	81.91 $\pm$ 24.62	44.51 $\pm$ 17.77	19.12 $\pm$ 10.40	5.31 $\pm$ 3.72	0.00 $\pm$ 0.00
	T6-8	89.66 $\pm$ 17.16	93.73 $\pm$ 17.67	108.02 $\pm$ 21.07	132.95 $\pm$ 24.29	139.89 $\pm$ 16.85	114.89 $\pm$ 31.55	71.25 $\pm$ 29.35	34.39 $\pm$ 23.20	10.93 $\pm$ 15.34	1.78 $\pm$ 5.95	0.00 $\pm$ 0.00
	T8-10	106.60 $\pm$ 19.51	110.54 $\pm$ 19.42	124.82 $\pm$ 22.14	137.97 $\pm$ 18.99	126.26 $\pm$ 26.35	89.10 $\pm$ 35.28	45.86 $\pm$ 32.93	12.33 $\pm$ 23.55	-5.39 $\pm$ 12.85	-4.52 $\pm$ 3.96	0.00 $\pm$ 0.00
	Global T1-10	78.25 $\pm$ 14.35	81.60 $\pm$ 15.02	95.05 $\pm$ 17.29	112.83 $\pm$ 14.97	119.10 $\pm$ 11.62	97.87 $\pm$ 22.55	59.87 $\pm$ 23.84	26.93 $\pm$ 16.51	6.55 $\pm$ 8.54	0.30 $\pm$ 2.60	0.00 $\pm$ 0.00
Lumbar	T10-L1	118.03 $\pm$ 16.21	122.16 $\pm$ 16.67	132.70 $\pm$ 16.51	131.50 $\pm$ 15.69	107.16 $\pm$ 26.34	66.80 $\pm$ 27.97	26.33 $\pm$ 25.09	-5.62 $\pm$ 20.08	-14.13 $\pm$ 9.54	-5.41 $\pm$ 3.15	0.00 $\pm$ 0.00
	L1-3	135.15 $\pm$ 12.45	138.67 $\pm$ 13.54	141.53 $\pm$ 17.79	128.13 $\pm$ 24.79	97.92 $\pm$ 29.42	60.20 $\pm$ 28.70	25.15 $\pm$ 26.41	-0.54 $\pm$ 20.47	-7.18 $\pm$ 10.69	-2.97 $\pm$ 3.76	0.00 $\pm$ 0.00
	L3-5	145.96 $\pm$ 14.61	145.36 $\pm$ 15.48	136.78 $\pm$ 18.63	114.62 $\pm$ 21.28	85.23 $\pm$ 20.94	54.51 $\pm$ 18.00	28.30 $\pm$ 14.20	9.28 $\pm$ 11.17	0.39 $\pm$ 8.38	-0.59 $\pm$ 3.02	0.00 $\pm$ 0.00
	Global T10-L5	133.44 $\pm$ 9.44	134.84 $\pm$ 9.52	134.72 $\pm$ 10.39	123.47 $\pm$ 13.92	97.59 $\pm$ 21.24	61.43 $\pm$ 20.70	27.89 $\pm$ 17.13	2.45 $\pm$ 11.77	-7.40 $\pm$ 6.57	-3.14 $\pm$ 2.58	0.00 $\pm$ 0.00

Normalize Segmental Difference $\pm$ SD for Pick object: Stoop (Begin cycle)

	Segment	Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	8.68 $\pm$ 8.62	16.98 $\pm$ 11.71	25.87 $\pm$ 14.28	28.61 $\pm$ 19.57	19.62 $\pm$ 24.27	-8.83 $\pm$ 21.16	-17.14 $\pm$ 16.87	-11.50 $\pm$ 14.45	-7.84 $\pm$ 17.05	-3.22 $\pm$ 12.54
	T4-6	0.00 $\pm$ 0.00	8.17 $\pm$ 9.34	18.81 $\pm$ 13.24	28.40 $\pm$ 15.84	32.17 $\pm$ 24.21	27.95 $\pm$ 26.11	1.53 $\pm$ 30.65	-18.93 $\pm$ 20.23	-15.27 $\pm$ 16.76	-10.68 $\pm$ 19.35	-3.35 $\pm$ 15.49
	T6-8	0.00 $\pm$ 0.00	5.23 $\pm$ 10.43	15.72 $\pm$ 15.60	29.19 $\pm$ 15.81	33.57 $\pm$ 20.96	32.73 $\pm$ 24.37	14.47 $\pm$ 29.61	-7.39 $\pm$ 24.60	-16.32 $\pm$ 18.43	-12.08 $\pm$ 18.38	-3.75 $\pm$ 15.21
	T8-10	0.00 $\pm$ 0.00	-1.09 $\pm$ 10.95	4.78 $\pm$ 20.57	23.87 $\pm$ 19.42	34.62 $\pm$ 16.47	34.73 $\pm$ 21.62	26.06 $\pm$ 24.56	8.14 $\pm$ 26.18	-11.20 $\pm$ 22.73	-11.05 $\pm$ 18.53	-3.52 $\pm$ 15.07
	Global T1-10	0.00 $\pm$ 0.00	3.66 $\pm$ 9.19	10.64 $\pm$ 14.51	25.09 $\pm$ 13.92	31.16 $\pm$ 16.40	26.61 $\pm$ 19.66	7.98 $\pm$ 17.74	-4.45 $\pm$ 15.61	-11.36 $\pm$ 16.09	-8.77 $\pm$ 16.18	-3.24 $\pm$ 12.70
Lumbar	T10-L1	0.00 $\pm$ 0.00	-4.85 $\pm$ 10.39	-1.65 $\pm$ 19.97	17.83 $\pm$ 21.53	33.54 $\pm$ 14.95	35.42 $\pm$ 17.18	29.44 $\pm$ 19.53	17.63 $\pm$ 20.33	-3.63 $\pm$ 22.47	-8.43 $\pm$ 19.35	-2.41 $\pm$ 16.59
	L1-3	0.00 $\pm$ 0.00	-3.07 $\pm$ 9.91	0.50 $\pm$ 17.51	14.65 $\pm$ 18.95	31.17 $\pm$ 11.70	32.38 $\pm$ 13.16	29.70 $\pm$ 14.74	23.85 $\pm$ 14.90	4.66 $\pm$ 21.51	-5.57 $\pm$ 18.72	-2.76 $\pm$ 14.05
	L3-5	0.00 $\pm$ 0.00	2.15 $\pm$ 7.47	7.06 $\pm$ 11.13	16.22 $\pm$ 12.86	23.21 $\pm$ 10.26	25.96 $\pm$ 8.58	25.75 $\pm$ 8.45	21.74 $\pm$ 13.04	12.49 $\pm$ 17.53	3.34 $\pm$ 15.32	-1.85 $\pm$ 12.48
	Global T10-L5	0.00 $\pm$ 0.00	-1.17 $\pm$ 8.22	2.47 $\pm$ 15.56	17.15 $\pm$ 16.78	28.83 $\pm$ 11.22	30.81 $\pm$ 11.70	27.77 $\pm$ 13.20	19.37 $\pm$ 14.95	4.66 $\pm$ 16.93	-2.85 $\pm$ 15.49	-1.48 $\pm$ 12.74

Normalize Segmental Difference $\pm$ SD for Pick object: Stoop (Return cycle)

	Segment	Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	2.79 $\pm$ 3.73	11.53 $\pm$ 7.04	21.92 $\pm$ 8.58	25.04 $\pm$ 12.41	-4.77 $\pm$ 28.13	-34.17 $\pm$ 11.03	-33.71 $\pm$ 6.95	-23.02 $\pm$ 7.22	-13.42 $\pm$ 5.08	-5.49 $\pm$ 2.79
	T4-6	0.00 $\pm$ 0.00	3.52 $\pm$ 2.78	13.79 $\pm$ 6.73	26.13 $\pm$ 8.54	23.34 $\pm$ 21.14	-12.48 $\pm$ 35.41	-40.38 $\pm$ 9.46	-37.40 $\pm$ 8.05	-25.39 $\pm$ 8.40	-13.81 $\pm$ 6.98	-5.31 $\pm$ 3.72
	T6-8	0.00 $\pm$ 0.00	4.07 $\pm$ 2.96	14.29 $\pm$ 6.99	24.92 $\pm$ 8.14	6.94 $\pm$ 30.07	-24.99 $\pm$ 31.34	-43.64 $\pm$ 7.04	-36.86 $\pm$ 8.89	-23.46 $\pm$ 11.04	-9.15 $\pm$ 10.29	-1.78 $\pm$ 5.95
	T8-10	0.00 $\pm$ 0.00	3.94 $\pm$ 3.26	14.28 $\pm$ 6.80	13.16 $\pm$ 17.71	-11.72 $\pm$ 30.37	-37.16 $\pm$ 16.47	-43.25 $\pm$ 6.51	-33.53 $\pm$ 13.54	-17.72 $\pm$ 15.64	0.87 $\pm$ 11.30	4.52 $\pm$ 3.96
	Global T1-10	0.00 $\pm$ 0.00	3.35 $\pm$ 3.43	13.45 $\pm$ 6.32	17.78 $\pm$ 9.08	6.26 $\pm$ 16.68	-21.22 $\pm$ 19.59	-38.00 $\pm$ 6.51	-32.94 $\pm$ 9.38	-20.39 $\pm$ 10.49	-6.25 $\pm$ 7.11	-0.30 $\pm$ 2.60
Lumbar	T10-L1	0.00 $\pm$ 0.00	4.13 $\pm$ 2.86	10.54 $\pm$ 9.33	-1.21 $\pm$ 19.57	-24.33 $\pm$ 21.29	-40.36 $\pm$ 5.81	-40.47 $\pm$ 6.44	-31.95 $\pm$ 10.42	-8.51 $\pm$ 16.89	8.71 $\pm$ 8.00	5.41 $\pm$ 3.15
	L1-3	0.00 $\pm$ 0.00	3.52 $\pm$ 3.84	2.86 $\pm$ 11.45	-13.40 $\pm$ 17.74	-30.21 $\pm$ 10.94	-37.72 $\pm$ 4.41	-35.05 $\pm$ 5.91	-25.69 $\pm$ 10.29	-6.64 $\pm$ 15.41	4.21 $\pm$ 7.80	2.97 $\pm$ 3.76
	L3-5	0.00 $\pm$ 0.00	-0.60 $\pm$ 3.88	-8.58 $\pm$ 8.08	-22.16 $\pm$ 8.23	-29.39 $\pm$ 4.45	-30.73 $\pm$ 5.13	-26.21 $\pm$ 6.19	-19.01 $\pm$ 6.55	-8.89 $\pm$ 6.96	-0.98 $\pm$ 5.66	0.59 $\pm$ 3.02
	Global T10-L5	0.00 $\pm$ 0.00	1.40 $\pm$ 2.49	-0.13 $\pm$ 5.40	-11.25 $\pm$ 11.83	-25.88 $\pm$ 13.63	-36.16 $\pm$ 4.14	-33.54 $\pm$ 5.82	-25.44 $\pm$ 8.49	-9.85 $\pm$ 10.81	4.26 $\pm$ 5.47	3.14 $\pm$ 2.58

Normalize Mean Difference $\pm$ SD for Pick object: Semi Squat (Begin cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	7.01 $\pm$ 4.13	23.40 $\pm$ 10.42	49.28 $\pm$ 17.05	77.06 $\pm$ 21.24	97.37 $\pm$ 23.43	102.75 $\pm$ 23.01	102.50 $\pm$ 22.79	101.52 $\pm$ 23.75	100.97 $\pm$ 24.88	100.21 $\pm$ 24.92
	T4-6	0.00 $\pm$ 0.00	6.88 $\pm$ 5.07	23.57 $\pm$ 12.89	51.33 $\pm$ 19.93	82.14 $\pm$ 24.94	110.09 $\pm$ 28.70	120.75 $\pm$ 25.47	122.76 $\pm$ 22.68	122.46 $\pm$ 22.06	121.17 $\pm$ 21.69	120.52 $\pm$ 21.22
	T6-8	0.00 $\pm$ 0.00	4.36 $\pm$ 5.38	18.76 $\pm$ 14.48	44.39 $\pm$ 24.62	74.68 $\pm$ 31.06	103.96 $\pm$ 34.91	122.38 $\pm$ 34.72	129.04 $\pm$ 30.02	131.43 $\pm$ 27.76	131.87 $\pm$ 27.45	131.28 $\pm$ 27.29
	T8-10	0.00 $\pm$ 0.00	-3.18 $\pm$ 5.41	2.01 $\pm$ 16.15	24.50 $\pm$ 26.72	52.53 $\pm$ 35.01	81.51 $\pm$ 38.83	103.61 $\pm$ 40.85	116.03 $\pm$ 38.76	121.85 $\pm$ 36.82	123.13 $\pm$ 36.31	122.52 $\pm$ 36.81
	Global T1-10	0.00 $\pm$ 0.00	1.59 $\pm$ 3.29	12.38 $\pm$ 11.66	36.44 $\pm$ 20.30	64.20 $\pm$ 27.06	87.94 $\pm$ 27.79	101.07 $\pm$ 25.98	107.71 $\pm$ 23.75	109.76 $\pm$ 22.02	109.59 $\pm$ 21.54	108.86 $\pm$ 21.62
Lumbar	T10-L1	0.00 $\pm$ 0.00	-5.58 $\pm$ 4.15	-9.70 $\pm$ 13.58	5.20 $\pm$ 24.83	32.09 $\pm$ 29.85	60.02 $\pm$ 31.66	81.28 $\pm$ 33.68	94.87 $\pm$ 34.35	101.41 $\pm$ 34.53	102.76 $\pm$ 35.23	101.92 $\pm$ 36.23
	L1-3	0.00 $\pm$ 0.00	-4.40 $\pm$ 5.08	-7.07 $\pm$ 12.16	5.72 $\pm$ 21.36	29.92 $\pm$ 25.72	54.61 $\pm$ 27.60	73.31 $\pm$ 29.87	85.78 $\pm$ 30.81	91.75 $\pm$ 30.97	93.00 $\pm$ 31.52	92.07 $\pm$ 31.96
	L3-5	0.00 $\pm$ 0.00	0.26 $\pm$ 3.07	4.99 $\pm$ 9.44	18.92 $\pm$ 15.54	37.70 $\pm$ 17.56	56.61 $\pm$ 18.67	70.90 $\pm$ 20.11	80.04 $\pm$ 21.57	83.89 $\pm$ 23.13	83.74 $\pm$ 25.40	82.59 $\pm$ 27.10
	Global T10-L5	0.00 $\pm$ 0.00	-2.49 $\pm$ 2.74	-2.31 $\pm$ 9.47	12.42 $\pm$ 18.16	34.89 $\pm$ 22.43	58.52 $\pm$ 23.79	76.40 $\pm$ 25.29	88.05 $\pm$ 26.58	93.55 $\pm$ 27.81	94.19 $\pm$ 29.22	93.05 $\pm$ 30.41

Normalize Mean Difference $\pm$ SD for Pick object: Semi Squat (Return cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	106.49 $\pm$ 21.09	107.30 $\pm$ 20.92	110.37 $\pm$ 20.19	110.84 $\pm$ 20.37	102.21 $\pm$ 24.23	80.28 $\pm$ 21.68	55.45 $\pm$ 16.72	32.11 $\pm$ 11.10	14.92 $\pm$ 5.87	4.55 $\pm$ 2.15	0.00 $\pm$ 0.00
	T4-6	124.64 $\pm$ 20.46	125.45 $\pm$ 20.29	127.29 $\pm$ 20.52	127.21 $\pm$ 24.95	115.29 $\pm$ 28.92	88.39 $\pm$ 22.88	60.31 $\pm$ 16.84	34.33 $\pm$ 11.74	15.58 $\pm$ 7.18	4.70 $\pm$ 2.60	0.00 $\pm$ 0.00
	T6-8	131.15 $\pm$ 27.64	130.83 $\pm$ 28.10	130.69 $\pm$ 29.81	124.86 $\pm$ 31.94	105.81 $\pm$ 29.89	78.55 $\pm$ 24.88	50.61 $\pm$ 21.00	20.74 $\pm$ 19.17	7.83 $\pm$ 12.11	0.97 $\pm$ 4.92	0.00 $\pm$ 0.00
	T8-10	119.84 $\pm$ 34.94	118.99 $\pm$ 35.62	115.10 $\pm$ 35.79	102.54 $\pm$ 32.58	82.13 $\pm$ 28.82	55.31 $\pm$ 24.88	28.06 $\pm$ 22.85	4.39 $\pm$ 18.92	-6.38 $\pm$ 11.74	-3.49 $\pm$ 4.02	0.00 $\pm$ 0.00
	Global T1-10	111.26 $\pm$ 19.97	111.12 $\pm$ 19.99	110.44 $\pm$ 20.93	104.63 $\pm$ 22.52	90.39 $\pm$ 22.21	67.10 $\pm$ 19.46	41.21 $\pm$ 15.36	18.01 $\pm$ 11.08	3.94 $\pm$ 6.39	0.41 $\pm$ 2.00	0.00 $\pm$ 0.00
Lumbar	T10-L1	102.41 $\pm$ 33.68	101.00 $\pm$ 33.80	95.82 $\pm$ 31.80	83.33 $\pm$ 27.76	63.69 $\pm$ 23.17	38.54 $\pm$ 19.38	13.00 $\pm$ 17.77	-8.35 $\pm$ 15.23	-11.57 $\pm$ 6.99	-4.93 $\pm$ 2.03	0.00 $\pm$ 0.00
	L1-3	92.90 $\pm$ 30.36	90.92 $\pm$ 29.81	85.94 $\pm$ 27.98	74.63 $\pm$ 24.61	56.85 $\pm$ 20.54	34.63 $\pm$ 17.45	12.61 $\pm$ 16.52	-5.23 $\pm$ 14.49	-8.55 $\pm$ 8.53	-3.53 $\pm$ 2.95	0.00 $\pm$ 0.00
	L3-5	81.12 $\pm$ 28.67	79.75 $\pm$ 28.26	75.98 $\pm$ 26.16	66.80 $\pm$ 23.01	52.52 $\pm$ 19.25	35.53 $\pm$ 15.39	19.74 $\pm$ 12.95	7.00 $\pm$ 11.52	1.73 $\pm$ 7.05	0.35 $\pm$ 2.62	0.00 $\pm$ 0.00
	Global T10-L5	91.86 $\pm$ 29.78	90.23 $\pm$ 29.67	85.65 $\pm$ 27.52	74.78 $\pm$ 23.74	57.62 $\pm$ 19.49	36.40 $\pm$ 15.60	15.61 $\pm$ 13.58	-1.41 $\pm$ 10.62	-6.00 $\pm$ 5.42	-2.32 $\pm$ 1.56	0.00 $\pm$ 0.00

Normalize Segmental Difference $\pm$ SD for Pick object: Semi Squat (Begin cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	7.01 $\pm$ 4.13	16.39 $\pm$ 6.65	25.88 $\pm$ 8.03	27.78 $\pm$ 7.61	20.31 $\pm$ 12.44	5.38 $\pm$ 15.89	-0.25 $\pm$ 11.44	-0.98 $\pm$ 7.21	-0.55 $\pm$ 4.43	-0.76 $\pm$ 3.65
	T4-6	0.00 $\pm$ 0.00	6.88 $\pm$ 5.07	16.69 $\pm$ 8.34	27.76 $\pm$ 8.87	30.81 $\pm$ 9.23	27.95 $\pm$ 9.95	10.66 $\pm$ 16.27	2.01 $\pm$ 13.56	-0.29 $\pm$ 9.23	-1.30 $\pm$ 6.11	-0.65 $\pm$ 4.56
	T6-8	0.00 $\pm$ 0.00	4.36 $\pm$ 5.38	14.41 $\pm$ 9.90	25.63 $\pm$ 12.02	30.29 $\pm$ 10.29	29.28 $\pm$ 10.15	18.42 $\pm$ 10.24	6.67 $\pm$ 13.30	2.38 $\pm$ 9.82	0.44 $\pm$ 5.66	-0.59 $\pm$ 4.81
	T8-10	0.00 $\pm$ 0.00	-3.18 $\pm$ 5.41	5.19 $\pm$ 13.25	22.49 $\pm$ 15.23	28.03 $\pm$ 14.18	28.98 $\pm$ 10.22	22.11 $\pm$ 8.96	12.42 $\pm$ 7.94	5.82 $\pm$ 7.69	1.29 $\pm$ 6.88	-0.62 $\pm$ 4.67
	Global T1-10	0.00 $\pm$ 0.00	1.59 $\pm$ 3.29	10.79 $\pm$ 9.31	24.06 $\pm$ 11.09	27.76 $\pm$ 10.83	23.74 $\pm$ 10.49	13.13 $\pm$ 8.27	6.64 $\pm$ 7.90	2.04 $\pm$ 6.29	-0.17 $\pm$ 4.15	-0.73 $\pm$ 3.44
Lumbar	T10-L1	0.00 $\pm$ 0.00	-5.58 $\pm$ 4.15	-4.12 $\pm$ 12.34	14.90 $\pm$ 18.42	26.89 $\pm$ 12.13	27.93 $\pm$ 8.71	21.25 $\pm$ 8.51	13.60 $\pm$ 7.06	6.54 $\pm$ 6.27	1.35 $\pm$ 6.31	-0.84 $\pm$ 4.72
	L1-3	0.00 $\pm$ 0.00	-4.40 $\pm$ 5.08	-2.67 $\pm$ 10.16	12.80 $\pm$ 14.93	24.20 $\pm$ 8.63	24.69 $\pm$ 7.22	18.70 $\pm$ 7.45	12.47 $\pm$ 6.54	5.97 $\pm$ 7.07	1.25 $\pm$ 7.00	-0.93 $\pm$ 4.26
	L3-5	0.00 $\pm$ 0.00	0.26 $\pm$ 3.07	4.73 $\pm$ 7.60	13.93 $\pm$ 8.06	18.77 $\pm$ 5.23	18.91 $\pm$ 6.53	14.29 $\pm$ 7.21	9.14 $\pm$ 6.84	3.85 $\pm$ 7.63	-0.15 $\pm$ 7.61	-1.15 $\pm$ 4.35
	Global T10-L5	0.00 $\pm$ 0.00	-2.49 $\pm$ 2.74	0.18 $\pm$ 8.92	14.73 $\pm$ 12.18	22.47 $\pm$ 8.99	23.63 $\pm$ 7.30	17.88 $\pm$ 7.67	11.65 $\pm$ 6.75	5.50 $\pm$ 6.88	0.64 $\pm$ 6.85	-1.14 $\pm$ 4.42

Normalize Segmental Difference $\pm$ SD for Pick object: Semi Squat (Return cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	0.81 $\pm$ 3.38	3.07 $\pm$ 5.74	0.48 $\pm$ 13.01	-8.63 $\pm$ 14.60	-21.93 $\pm$ 11.04	-24.83 $\pm$ 7.37	-23.34 $\pm$ 6.75	-17.19 $\pm$ 5.91	-10.37 $\pm$ 4.02	-4.55 $\pm$ 2.15
	T4-6	0.00 $\pm$ 0.00	0.81 $\pm$ 3.69	1.84 $\pm$ 6.94	-0.09 $\pm$ 13.82	-11.92 $\pm$ 14.73	-26.90 $\pm$ 8.53	-28.08 $\pm$ 8.00	-25.98 $\pm$ 6.89	-18.75 $\pm$ 5.65	-10.89 $\pm$ 5.00	-4.70 $\pm$ 2.60
	T6-8	0.00 $\pm$ 0.00	-0.32 $\pm$ 2.94	-0.14 $\pm$ 7.85	-5.83 $\pm$ 11.25	-19.05 $\pm$ 10.50	-27.26 $\pm$ 8.76	-27.94 $\pm$ 8.08	-29.87 $\pm$ 16.32	-12.92 $\pm$ 14.88	-6.85 $\pm$ 8.23	-0.97 $\pm$ 4.92
	T8-10	0.00 $\pm$ 0.00	-0.84 $\pm$ 3.74	-3.89 $\pm$ 6.35	-12.57 $\pm$ 7.26	-20.41 $\pm$ 8.02	-26.81 $\pm$ 8.47	-27.26 $\pm$ 7.78	-23.67 $\pm$ 9.76	-10.77 $\pm$ 11.60	2.89 $\pm$ 8.65	3.49 $\pm$ 4.02
	Global T1-10	0.00 $\pm$ 0.00	-0.13 $\pm$ 3.27	-0.69 $\pm$ 5.40	-5.81 $\pm$ 6.37	-14.24 $\pm$ 7.31	-23.30 $\pm$ 8.93	-25.89 $\pm$ 7.62	-23.20 $\pm$ 7.63	-14.06 $\pm$ 7.33	-3.53 $\pm$ 4.95	-0.41 $\pm$ 2.00
Lumbar	T10-L1	0.00 $\pm$ 0.00	-1.41 $\pm$ 2.80	-5.19 $\pm$ 4.77	-12.49 $\pm$ 6.92	-19.64 $\pm$ 7.59	-25.16 $\pm$ 7.80	-25.54 $\pm$ 6.97	-21.35 $\pm$ 8.79	-3.21 $\pm$ 11.91	6.63 $\pm$ 6.10	4.93 $\pm$ 2.03
	L1-3	0.00 $\pm$ 0.00	-1.97 $\pm$ 2.90	-4.99 $\pm$ 4.91	-11.31 $\pm$ 6.23	-17.78 $\pm$ 6.66	-22.22 $\pm$ 6.52	-22.03 $\pm$ 5.78	-17.83 $\pm$ 8.49	-3.33 $\pm$ 9.36	5.03 $\pm$ 6.20	3.53 $\pm$ 2.95
	L3-5	0.00 $\pm$ 0.00	-1.37 $\pm$ 2.55	-3.77 $\pm$ 5.08	-9.18 $\pm$ 6.57	-14.28 $\pm$ 6.57	-16.98 $\pm$ 6.45	-15.80 $\pm$ 5.61	-12.74 $\pm$ 5.14	-5.27 $\pm$ 5.60	-1.39 $\pm$ 4.91	-0.35 $\pm$ 2.62
	Global T10-L5	0.00 $\pm$ 0.00	-1.63 $\pm$ 2.53	-4.58 $\pm$ 4.85	-10.88 $\pm$ 6.60	-17.15 $\pm$ 6.90	-21.22 $\pm$ 6.80	-20.79 $\pm$ 5.71	-17.02 $\pm$ 6.76	-4.59 $\pm$ 8.01	3.68 $\pm$ 4.48	2.32 $\pm$ 1.56

Normalize Mean Difference $\pm$ SD for Pick object: Squat (Begin cycle)

	Segment	Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	5.61 $\pm$ 3.35	17.77 $\pm$ 9.23	34.99 $\pm$ 17.75	51.55 $\pm$ 25.86	62.87 $\pm$ 29.91	68.53 $\pm$ 31.28	70.61 $\pm$ 31.35	70.53 $\pm$ 29.90	69.04 $\pm$ 27.93	67.92 $\pm$ 27.34
	T4-6	0.00 $\pm$ 0.00	5.33 $\pm$ 3.83	17.90 $\pm$ 10.53	36.83 $\pm$ 19.73	55.76 $\pm$ 27.90	69.89 $\pm$ 33.39	78.09 $\pm$ 35.28	82.30 $\pm$ 35.36	83.60 $\pm$ 33.80	82.69 $\pm$ 31.69	81.45 $\pm$ 30.30
	T6-8	0.00 $\pm$ 0.00	3.02 $\pm$ 4.66	12.64 $\pm$ 13.05	30.53 $\pm$ 22.41	49.49 $\pm$ 30.78	64.30 $\pm$ 35.63	73.43 $\pm$ 37.69	78.25 $\pm$ 37.64	79.51 $\pm$ 35.31	78.95 $\pm$ 34.04	78.22 $\pm$ 34.31
	T8-10	0.00 $\pm$ 0.00	-3.81 $\pm$ 4.31	-3.93 $\pm$ 12.39	8.37 $\pm$ 23.75	25.50 $\pm$ 32.32	39.98 $\pm$ 37.54	49.10 $\pm$ 39.68	54.06 $\pm$ 39.52	55.49 $\pm$ 38.06	54.46 $\pm$ 36.57	53.29 $\pm$ 35.51
	Global T1-10	0.00 $\pm$ 0.00	0.74 $\pm$ 2.30	6.47 $\pm$ 8.80	21.22 $\pm$ 18.45	37.66 $\pm$ 26.91	50.36 $\pm$ 31.08	57.38 $\pm$ 32.42	60.51 $\pm$ 31.80	61.18 $\pm$ 30.44	60.08 $\pm$ 29.27	58.96 $\pm$ 28.77
Lumbar	T10-L1	0.00 $\pm$ 0.00	-5.20 $\pm$ 4.01	-10.72 $\pm$ 11.02	-4.43 $\pm$ 21.22	10.79 $\pm$ 29.09	24.92 $\pm$ 32.71	33.75 $\pm$ 33.67	38.06 $\pm$ 32.10	38.84 $\pm$ 29.92	37.31 $\pm$ 27.84	36.17 $\pm$ 26.46
	L1-3	0.00 $\pm$ 0.00	-3.37 $\pm$ 5.15	-7.71 $\pm$ 11.61	0.55 $\pm$ 18.73	15.87 $\pm$ 23.69	28.67 $\pm$ 25.57	35.99 $\pm$ 25.68	39.10 $\pm$ 24.57	38.34 $\pm$ 23.52	35.01 $\pm$ 22.64	32.53 $\pm$ 22.08
	L3-5	0.00 $\pm$ 0.00	0.67 $\pm$ 2.93	5.41 $\pm$ 8.18	18.36 $\pm$ 11.02	31.61 $\pm$ 13.66	40.37 $\pm$ 15.56	44.06 $\pm$ 15.84	43.62 $\pm$ 16.49	39.40 $\pm$ 17.23	33.21 $\pm$ 17.86	29.16 $\pm$ 18.58
	Global T10-L5	0.00 $\pm$ 0.00	-2.28 $\pm$ 2.71	-2.35 $\pm$ 7.94	7.94 $\pm$ 15.25	22.19 $\pm$ 21.29	33.39 $\pm$ 24.26	39.47 $\pm$ 24.63	41.47 $\pm$ 23.59	39.83 $\pm$ 22.25	35.90 $\pm$ 21.35	33.05 $\pm$ 20.98

Normalize Mean Difference $\pm$ SD for Pick object: Squat (Return cycle)

	Segment	Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	72.47 $\pm$ 24.17	72.46 $\pm$ 24.45	73.15 $\pm$ 23.77	71.31 $\pm$ 23.48	64.52 $\pm$ 23.24	52.49 $\pm$ 20.70	38.63 $\pm$ 17.96	23.96 $\pm$ 13.47	11.98 $\pm$ 8.86	3.96 $\pm$ 4.20	0.00 $\pm$ 0.00
	T4-6	82.78 $\pm$ 28.57	83.00 $\pm$ 29.06	84.35 $\pm$ 29.64	82.23 $\pm$ 29.48	73.41 $\pm$ 27.34	59.43 $\pm$ 24.65	43.11 $\pm$ 20.79	26.77 $\pm$ 15.89	13.55 $\pm$ 10.01	4.45 $\pm$ 4.54	0.00 $\pm$ 0.00
	T6-8	76.29 $\pm$ 33.29	75.84 $\pm$ 32.80	76.85 $\pm$ 32.20	75.08 $\pm$ 32.37	66.33 $\pm$ 31.15	51.76 $\pm$ 28.47	35.06 $\pm$ 25.37	18.56 $\pm$ 21.03	7.27 $\pm$ 13.75	1.84 $\pm$ 6.14	0.00 $\pm$ 0.00
	T8-10	54.65 $\pm$ 35.42	54.94 $\pm$ 35.32	56.68 $\pm$ 35.32	54.70 $\pm$ 34.84	45.99 $\pm$ 33.86	31.74 $\pm$ 32.00	15.77 $\pm$ 28.57	2.01 $\pm$ 22.35	-3.81 $\pm$ 13.70	-2.07 $\pm$ 5.26	0.00 $\pm$ 0.00
	Global T1-10	62.18 $\pm$ 27.05	62.32 $\pm$ 26.96	63.41 $\pm$ 26.43	61.50 $\pm$ 25.95	53.89 $\pm$ 25.41	41.01 $\pm$ 23.64	26.49 $\pm$ 21.54	12.45 $\pm$ 16.50	3.88 $\pm$ 10.30	0.69 $\pm$ 4.62	0.00 $\pm$ 0.00
Lumbar	T10-L1	39.92 $\pm$ 26.32	40.43 $\pm$ 26.40	42.86 $\pm$ 26.85	41.98 $\pm$ 27.77	34.11 $\pm$ 28.60	20.51 $\pm$ 28.59	5.91 $\pm$ 25.73	-5.08 $\pm$ 18.94	-8.08 $\pm$ 11.10	-3.60 $\pm$ 3.66	0.00 $\pm$ 0.00
	L1-3	35.25 $\pm$ 21.03	36.47 $\pm$ 20.49	40.51 $\pm$ 19.74	41.25 $\pm$ 19.93	34.89 $\pm$ 21.06	23.09 $\pm$ 21.66	9.50 $\pm$ 20.93	-1.31 $\pm$ 16.59	-5.55 $\pm$ 10.53	-2.40 $\pm$ 3.57	0.00 $\pm$ 0.00
	L3-5	27.12 $\pm$ 18.64	29.57 $\pm$ 17.89	36.37 $\pm$ 16.18	40.43 $\pm$ 13.80	37.55 $\pm$ 12.58	29.65 $\pm$ 12.07	19.87 $\pm$ 11.25	9.92 $\pm$ 10.12	3.25 $\pm$ 7.39	0.73 $\pm$ 2.70	0.00 $\pm$ 0.00
	Global T10-L5	33.42 $\pm$ 20.96	35.02 $\pm$ 20.84	39.92 $\pm$ 20.39	41.75 $\pm$ 19.82	36.51 $\pm$ 19.65	25.77 $\pm$ 19.25	13.34 $\pm$ 17.46	2.22 $\pm$ 13.72	-2.81 $\pm$ 8.87	-1.38 $\pm$ 3.18	0.00 $\pm$ 0.00

Normalize Segmental Difference $\pm$ SD for Pick object: Squat (Begin cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	5.61 $\pm$ 3.35	12.16 $\pm$ 6.44	17.22 $\pm$ 9.60	16.57 $\pm$ 9.39	11.31 $\pm$ 6.55	5.66 $\pm$ 4.23	2.09 $\pm$ 3.44	-0.08 $\pm$ 4.68	-1.49 $\pm$ 5.06	-1.12 $\pm$ 4.61
	T4-6	0.00 $\pm$ 0.00	5.33 $\pm$ 3.83	12.57 $\pm$ 7.20	18.93 $\pm$ 10.38	18.93 $\pm$ 9.92	14.14 $\pm$ 7.56	8.20 $\pm$ 5.62	4.20 $\pm$ 4.21	1.30 $\pm$ 5.22	-0.91 $\pm$ 5.70	-1.24 $\pm$ 5.47
	T6-8	0.00 $\pm$ 0.00	3.02 $\pm$ 4.66	9.62 $\pm$ 9.22	17.89 $\pm$ 11.23	18.96 $\pm$ 10.54	14.81 $\pm$ 7.37	9.13 $\pm$ 5.85	4.82 $\pm$ 5.18	1.26 $\pm$ 6.01	-0.56 $\pm$ 5.73	-0.72 $\pm$ 5.30
	T8-10	0.00 $\pm$ 0.00	-3.81 $\pm$ 4.31	-0.12 $\pm$ 10.65	12.30 $\pm$ 15.58	17.12 $\pm$ 12.13	14.49 $\pm$ 8.05	9.12 $\pm$ 6.08	4.96 $\pm$ 5.26	1.43 $\pm$ 5.98	-1.03 $\pm$ 5.47	-1.18 $\pm$ 5.15
	Global T1-10	0.00 $\pm$ 0.00	0.74 $\pm$ 2.30	5.74 $\pm$ 7.44	14.75 $\pm$ 11.68	16.44 $\pm$ 10.53	12.71 $\pm$ 6.46	7.02 $\pm$ 4.60	3.12 $\pm$ 4.10	0.68 $\pm$ 5.08	-1.11 $\pm$ 4.68	-1.12 $\pm$ 4.51
Lumbar	T10-L1	0.00 $\pm$ 0.00	-5.20 $\pm$ 4.01	-5.51 $\pm$ 9.30	6.29 $\pm$ 16.62	15.22 $\pm$ 12.42	14.13 $\pm$ 7.49	8.83 $\pm$ 5.53	4.31 $\pm$ 5.99	0.79 $\pm$ 6.17	-1.53 $\pm$ 5.37	-1.14 $\pm$ 4.61
	L1-3	0.00 $\pm$ 0.00	-3.37 $\pm$ 5.15	-4.34 $\pm$ 7.82	8.26 $\pm$ 14.13	15.33 $\pm$ 9.11	12.80 $\pm$ 6.11	7.32 $\pm$ 5.17	3.11 $\pm$ 5.98	-0.76 $\pm$ 5.97	-3.34 $\pm$ 4.67	-2.47 $\pm$ 3.86
	L3-5	0.00 $\pm$ 0.00	0.67 $\pm$ 2.93	4.74 $\pm$ 6.37	12.95 $\pm$ 6.14	13.25 $\pm$ 6.56	8.76 $\pm$ 5.55	3.69 $\pm$ 5.20	-0.44 $\pm$ 5.66	-4.22 $\pm$ 5.16	-6.20 $\pm$ 4.80	-4.05 $\pm$ 3.42
	Global T10-L5	0.00 $\pm$ 0.00	-2.28 $\pm$ 2.71	-0.07 $\pm$ 7.33	10.28 $\pm$ 11.24	14.25 $\pm$ 9.33	11.20 $\pm$ 6.11	6.08 $\pm$ 5.18	2.01 $\pm$ 5.28	-1.64 $\pm$ 5.18	-3.93 $\pm$ 4.42	-2.85 $\pm$ 3.40

Normalize Segmental Difference $\pm$ SD for Pick object: Squat (Return cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	0.00 $\pm$ 3.18	0.69 $\pm$ 4.25	-1.84 $\pm$ 4.68	-6.79 $\pm$ 4.80	-12.03 $\pm$ 6.33	-13.86 $\pm$ 5.46	-14.67 $\pm$ 5.41	-11.98 $\pm$ 5.32	-8.02 $\pm$ 4.96	-3.96 $\pm$ 4.20
	T4-6	0.00 $\pm$ 0.00	0.22 $\pm$ 4.05	1.35 $\pm$ 5.22	-2.12 $\pm$ 5.12	-8.81 $\pm$ 5.60	-13.98 $\pm$ 5.77	-16.32 $\pm$ 5.84	-16.33 $\pm$ 6.09	-13.22 $\pm$ 6.53	-9.10 $\pm$ 5.80	-4.45 $\pm$ 4.54
	T6-8	0.00 $\pm$ 0.00	-0.45 $\pm$ 3.32	1.01 $\pm$ 4.85	-1.77 $\pm$ 5.74	-8.76 $\pm$ 5.53	-14.57 $\pm$ 6.07	-16.70 $\pm$ 5.49	-16.49 $\pm$ 6.19	-11.30 $\pm$ 8.60	-5.43 $\pm$ 8.06	-1.84 $\pm$ 6.14
	T8-10	0.00 $\pm$ 0.00	0.28 $\pm$ 3.45	1.74 $\pm$ 5.08	-1.97 $\pm$ 5.55	-8.72 $\pm$ 5.58	-14.24 $\pm$ 5.79	-15.97 $\pm$ 6.42	-13.76 $\pm$ 9.93	-5.82 $\pm$ 11.63	1.74 $\pm$ 9.23	2.07 $\pm$ 5.26
	Global T1-10	0.00 $\pm$ 0.00	0.14 $\pm$ 2.86	1.09 $\pm$ 4.26	-1.92 $\pm$ 4.85	-7.61 $\pm$ 4.95	-12.88 $\pm$ 5.77	-14.52 $\pm$ 5.66	-14.03 $\pm$ 7.02	-8.57 $\pm$ 7.82	-3.19 $\pm$ 6.34	-0.69 $\pm$ 4.62
Lumbar	T10-L1	0.00 $\pm$ 0.00	0.51 $\pm$ 3.23	2.43 $\pm$ 4.96	-0.88 $\pm$ 5.65	-7.87 $\pm$ 5.42	-13.60 $\pm$ 5.34	-14.60 $\pm$ 6.49	-10.99 $\pm$ 11.78	-3.00 $\pm$ 11.29	4.48 $\pm$ 8.07	3.60 $\pm$ 3.66
	L1-3	0.00 $\pm$ 0.00	1.21 $\pm$ 3.32	4.05 $\pm$ 5.30	0.73 $\pm$ 5.94	-6.36 $\pm$ 5.47	-11.80 $\pm$ 4.84	-13.59 $\pm$ 4.45	-10.81 $\pm$ 8.57	-4.23 $\pm$ 9.08	3.15 $\pm$ 7.48	2.40 $\pm$ 3.57
	L3-5	0.00 $\pm$ 0.00	2.45 $\pm$ 3.46	6.80 $\pm$ 5.29	4.06 $\pm$ 5.86	-2.88 $\pm$ 5.61	-7.70 $\pm$ 4.68	-9.78 $\pm$ 3.61	-9.95 $\pm$ 3.70	-6.67 $\pm$ 4.21	-2.52 $\pm$ 4.92	-0.73 $\pm$ 2.70
	Global T10-L5	0.00 $\pm$ 0.00	1.59 $\pm$ 3.11	4.90 $\pm$ 4.59	1.84 $\pm$ 5.15	-5.25 $\pm$ 5.21	-10.47 $\pm$ 4.96	-12.43 $\pm$ 4.42	-11.12 $\pm$ 7.01	-5.03 $\pm$ 7.40	1.43 $\pm$ 6.03	1.38 $\pm$ 3.18

Appendix K Normalize Segmental and Mean Differences during Weightlift of 5kg load using different approaches.

Normalize Mean Difference $\pm$ SD for Weightlift 5kg: Stoop (Begin cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	5.39 $\pm$ 3.34	19.02 $\pm$ 7.90	43.85 $\pm$ 14.45	76.26 $\pm$ 19.51	99.98 $\pm$ 15.92	106.39 $\pm$ 23.82	97.71 $\pm$ 27.40	88.46 $\pm$ 23.72	82.93 $\pm$ 22.44	80.88 $\pm$ 22.43
	T4-6	0.00 $\pm$ 0.00	4.83 $\pm$ 3.03	18.28 $\pm$ 8.50	44.30 $\pm$ 16.32	79.25 $\pm$ 22.69	113.29 $\pm$ 20.53	132.95 $\pm$ 16.64	132.20 $\pm$ 22.41	123.04 $\pm$ 24.83	116.19 $\pm$ 25.20	113.88 $\pm$ 25.64
	T6-8	0.00 $\pm$ 0.00	1.99 $\pm$ 3.95	11.35 $\pm$ 11.95	35.89 $\pm$ 20.90	70.73 $\pm$ 26.61	108.60 $\pm$ 28.42	135.65 $\pm$ 19.39	149.59 $\pm$ 13.48	150.05 $\pm$ 14.37	146.35 $\pm$ 15.46	144.33 $\pm$ 15.77
	T8-10	0.00 $\pm$ 0.00	-2.89 $\pm$ 3.89	-4.03 $\pm$ 14.05	12.53 $\pm$ 27.33	45.50 $\pm$ 34.08	83.09 $\pm$ 34.99	115.43 $\pm$ 30.89	137.41 $\pm$ 20.83	148.22 $\pm$ 13.51	150.74 $\pm$ 11.22	150.31 $\pm$ 11.65
	Global T1-10	0.00 $\pm$ 0.00	1.07 $\pm$ 2.43	7.20 $\pm$ 9.08	28.01 $\pm$ 18.83	59.98 $\pm$ 23.77	89.53 $\pm$ 17.37	108.94 $\pm$ 11.24	114.78 $\pm$ 11.85	115.54 $\pm$ 12.46	114.19 $\pm$ 12.32	112.91 $\pm$ 12.43
Lumbar	T10-L1	0.00 $\pm$ 0.00	-4.05 $\pm$ 2.75	-10.77 $\pm$ 10.19	-1.73 $\pm$ 22.57	27.49 $\pm$ 30.93	63.13 $\pm$ 31.62	94.81 $\pm$ 28.22	119.17 $\pm$ 21.95	133.14 $\pm$ 16.83	139.05 $\pm$ 14.04	140.54 $\pm$ 13.40
	L1-3	0.00 $\pm$ 0.00	-2.38 $\pm$ 2.96	-6.30 $\pm$ 10.84	1.20 $\pm$ 22.02	28.10 $\pm$ 26.35	60.16 $\pm$ 26.90	88.83 $\pm$ 24.85	110.87 $\pm$ 20.87	124.05 $\pm$ 18.49	130.51 $\pm$ 18.14	132.39 $\pm$ 18.01
	L3-5	0.00 $\pm$ 0.00	0.30 $\pm$ 2.61	4.16 $\pm$ 7.80	16.70 $\pm$ 14.65	37.52 $\pm$ 19.92	62.10 $\pm$ 23.08	85.21 $\pm$ 23.76	104.08 $\pm$ 22.66	115.89 $\pm$ 22.01	121.81 $\pm$ 22.17	123.59 $\pm$ 22.27
	Global T10-L5	0.00 $\pm$ 0.00	-1.99 $\pm$ 2.12	-3.66 $\pm$ 7.59	6.31 $\pm$ 16.85	31.82 $\pm$ 22.84	62.12 $\pm$ 24.58	89.83 $\pm$ 22.86	111.44 $\pm$ 18.31	124.35 $\pm$ 15.18	130.59 $\pm$ 14.72	132.35 $\pm$ 14.60

Normalize Mean Difference $\pm$ SD for Weightlift 5kg: Stoop (Return cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	93.49 $\pm$ 25.21	95.96 $\pm$ 25.41	103.84 $\pm$ 27.87	117.23 $\pm$ 26.77	117.61 $\pm$ 19.46	93.78 $\pm$ 18.21	60.24 $\pm$ 14.48	32.16 $\pm$ 9.71	13.98 $\pm$ 5.36	3.88 $\pm$ 2.01	0.00 $\pm$ 0.00
	T4-6	120.96 $\pm$ 26.22	123.76 $\pm$ 25.95	133.28 $\pm$ 25.38	143.61 $\pm$ 20.29	132.92 $\pm$ 16.75	98.88 $\pm$ 19.34	62.30 $\pm$ 16.97	31.71 $\pm$ 13.68	12.40 $\pm$ 8.27	3.07 $\pm$ 2.98	0.00 $\pm$ 0.00
	T6-8	143.18 $\pm$ 13.60	145.29 $\pm$ 13.22	150.46 $\pm$ 11.55	147.07 $\pm$ 15.85	121.55 $\pm$ 24.12	83.48 $\pm$ 24.81	46.85 $\pm$ 23.37	17.10 $\pm$ 19.37	1.33 $\pm$ 12.46	-1.03 $\pm$ 3.75	0.00 $\pm$ 0.00
	T8-10	145.75 $\pm$ 11.46	146.46 $\pm$ 10.71	143.53 $\pm$ 13.99	125.18 $\pm$ 19.69	93.95 $\pm$ 22.63	56.67 $\pm$ 23.96	21.17 $\pm$ 23.45	-6.01 $\pm$ 20.55	-9.40 $\pm$ 9.49	-3.68 $\pm$ 2.63	0.00 $\pm$ 0.00
	Global T1-10	116.22 $\pm$ 13.03	117.82 $\pm$ 12.83	120.48 $\pm$ 12.81	118.06 $\pm$ 11.07	103.30 $\pm$ 12.86	73.36 $\pm$ 16.58	39.39 $\pm$ 15.46	12.25 $\pm$ 11.86	1.67 $\pm$ 5.46	0.03 $\pm$ 1.39	0.00 $\pm$ 0.00
Lumbar	T10-L1	138.04 $\pm$ 13.59	135.99 $\pm$ 14.91	127.39 $\pm$ 16.67	106.78 $\pm$ 18.28	76.94 $\pm$ 20.67	41.74 $\pm$ 22.15	8.69 $\pm$ 21.88	-13.83 $\pm$ 15.57	-11.97 $\pm$ 6.13	-3.96 $\pm$ 2.23	0.00 $\pm$ 0.00
	L1-3	131.75 $\pm$ 18.74	129.30 $\pm$ 19.40	119.53 $\pm$ 21.18	99.49 $\pm$ 22.55	72.17 $\pm$ 23.12	41.03 $\pm$ 22.56	12.51 $\pm$ 21.49	-6.39 $\pm$ 14.43	-8.28 $\pm$ 7.55	-2.77 $\pm$ 2.44	0.00 $\pm$ 0.00
	L3-5	122.77 $\pm$ 21.85	119.73 $\pm$ 22.25	110.06 $\pm$ 22.74	92.16 $\pm$ 22.70	68.98 $\pm$ 21.11	44.11 $\pm$ 17.86	22.74 $\pm$ 14.52	7.37 $\pm$ 11.23	0.64 $\pm$ 6.80	-0.34 $\pm$ 2.50	0.00 $\pm$ 0.00
	Global T10-L5	130.24 $\pm$ 13.54	127.85 $\pm$ 14.43	118.57 $\pm$ 15.58	98.95 $\pm$ 16.40	72.00 $\pm$ 16.79	41.67 $\pm$ 15.85	14.28 $\pm$ 14.17	-5.29 $\pm$ 9.75	-6.45 $\pm$ 4.81	-2.07 $\pm$ 1.53	0.00 $\pm$ 0.00

Normalize Segmental Difference $\pm$ SD for Weightlift 5kg: Stoop (Begin cycle)

	Segment	Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	5.39 $\pm$ 3.34	13.63 $\pm$ 5.01	24.82 $\pm$ 7.59	32.41 $\pm$ 7.27	23.72 $\pm$ 18.61	6.41 $\pm$ 21.39	-8.68 $\pm$ 14.28	-9.25 $\pm$ 7.11	-5.53 $\pm$ 5.92	-2.05 $\pm$ 4.25
	T4-6	0.00 $\pm$ 0.00	4.83 $\pm$ 3.03	13.46 $\pm$ 5.91	26.02 $\pm$ 9.46	34.95 $\pm$ 8.14	34.05 $\pm$ 8.73	19.66 $\pm$ 21.85	-0.75 $\pm$ 20.49	-9.16 $\pm$ 8.49	-6.85 $\pm$ 4.23	-2.31 $\pm$ 2.27
	T6-8	0.00 $\pm$ 0.00	1.99 $\pm$ 3.95	9.37 $\pm$ 8.48	24.54 $\pm$ 11.38	34.83 $\pm$ 8.11	37.87 $\pm$ 5.47	27.05 $\pm$ 14.89	13.94 $\pm$ 17.55	0.46 $\pm$ 12.41	-3.70 $\pm$ 6.93	-2.02 $\pm$ 2.15
	T8-10	0.00 $\pm$ 0.00	-2.89 $\pm$ 3.89	-1.15 $\pm$ 11.56	16.56 $\pm$ 17.60	32.97 $\pm$ 10.79	37.59 $\pm$ 5.51	32.34 $\pm$ 6.18	21.98 $\pm$ 12.44	10.80 $\pm$ 10.07	2.53 $\pm$ 6.66	-0.44 $\pm$ 4.41
	Global T1-10	0.00 $\pm$ 0.00	1.07 $\pm$ 2.43	6.12 $\pm$ 7.40	20.82 $\pm$ 11.66	31.97 $\pm$ 8.41	29.55 $\pm$ 10.53	19.41 $\pm$ 12.03	5.84 $\pm$ 9.81	0.76 $\pm$ 3.73	-1.34 $\pm$ 4.40	-1.28 $\pm$ 2.75
Lumbar	T10-L1	0.00 $\pm$ 0.00	-4.05 $\pm$ 2.75	-6.72 $\pm$ 9.02	9.04 $\pm$ 16.96	29.23 $\pm$ 13.10	35.64 $\pm$ 5.13	31.68 $\pm$ 5.43	24.36 $\pm$ 8.07	13.98 $\pm$ 7.98	5.91 $\pm$ 5.58	1.50 $\pm$ 2.10
	L1-3	0.00 $\pm$ 0.00	-2.38 $\pm$ 2.96	-3.92 $\pm$ 8.84	7.49 $\pm$ 16.66	26.90 $\pm$ 9.91	32.06 $\pm$ 6.43	28.66 $\pm$ 6.01	22.05 $\pm$ 7.59	13.18 $\pm$ 7.16	6.47 $\pm$ 4.14	1.87 $\pm$ 1.91
	L3-5	0.00 $\pm$ 0.00	0.30 $\pm$ 2.61	3.86 $\pm$ 5.71	12.54 $\pm$ 8.49	20.82 $\pm$ 7.36	24.58 $\pm$ 6.32	23.11 $\pm$ 5.99	18.87 $\pm$ 6.20	11.81 $\pm$ 5.54	5.92 $\pm$ 3.58	1.78 $\pm$ 1.43
	Global T10-L5	0.00 $\pm$ 0.00	-1.99 $\pm$ 2.12	-1.67 $\pm$ 6.84	9.97 $\pm$ 11.82	25.52 $\pm$ 8.72	30.30 $\pm$ 5.39	27.71 $\pm$ 5.46	21.61 $\pm$ 7.00	12.91 $\pm$ 6.97	6.25 $\pm$ 3.87	1.75 $\pm$ 1.65

Normalize Segmental Difference $\pm$ SD for Weightlift 5kg: Stoop (Return cycle)

	Segment	Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	2.48 $\pm$ 4.43	7.88 $\pm$ 6.65	13.39 $\pm$ 11.31	0.38 $\pm$ 22.15	-23.83 $\pm$ 18.24	-33.54 $\pm$ 6.65	-28.07 $\pm$ 6.23	-18.19 $\pm$ 5.36	-10.09 $\pm$ 3.67	-3.88 $\pm$ 2.01
	T4-6	0.00 $\pm$ 0.00	2.80 $\pm$ 1.98	9.52 $\pm$ 5.56	10.33 $\pm$ 13.90	-10.69 $\pm$ 24.23	-34.04 $\pm$ 8.11	-36.58 $\pm$ 5.53	-30.59 $\pm$ 6.38	-19.31 $\pm$ 6.97	-9.33 $\pm$ 5.58	-3.07 $\pm$ 2.98
	T6-8	0.00 $\pm$ 0.00	2.11 $\pm$ 2.58	5.16 $\pm$ 8.24	-3.39 $\pm$ 16.87	-25.51 $\pm$ 13.54	-38.07 $\pm$ 4.63	-36.63 $\pm$ 5.62	-29.76 $\pm$ 8.50	-15.77 $\pm$ 8.90	-2.35 $\pm$ 9.43	1.03 $\pm$ 3.75
	T8-10	0.00 $\pm$ 0.00	0.71 $\pm$ 3.36	-2.93 $\pm$ 8.82	-18.35 $\pm$ 10.52	-31.23 $\pm$ 5.07	-37.28 $\pm$ 4.24	-35.49 $\pm$ 5.62	-27.18 $\pm$ 9.87	-3.39 $\pm$ 14.72	5.72 $\pm$ 7.88	3.68 $\pm$ 2.63
	Global T1-10	0.00 $\pm$ 0.00	1.60 $\pm$ 3.35	2.66 $\pm$ 5.50	-2.43 $\pm$ 7.96	-14.76 $\pm$ 12.82	-29.94 $\pm$ 10.46	-33.97 $\pm$ 6.54	-27.14 $\pm$ 7.75	-10.57 $\pm$ 8.24	-1.64 $\pm$ 4.51	-0.03 $\pm$ 1.39
Lumbar	T10-L1	0.00 $\pm$ 0.00	-2.05 $\pm$ 2.89	-8.60 $\pm$ 5.38	-20.61 $\pm$ 6.83	-29.84 $\pm$ 5.02	-35.20 $\pm$ 4.77	-33.06 $\pm$ 5.11	-22.51 $\pm$ 13.01	1.85 $\pm$ 13.37	8.01 $\pm$ 4.92	3.96 $\pm$ 2.23
	L1-3	0.00 $\pm$ 0.00	-2.45 $\pm$ 2.36	-9.77 $\pm$ 4.62	-20.03 $\pm$ 5.73	-27.32 $\pm$ 4.98	-31.14 $\pm$ 4.98	-28.52 $\pm$ 5.09	-18.90 $\pm$ 12.04	-1.89 $\pm$ 9.67	5.50 $\pm$ 5.62	2.77 $\pm$ 2.44
	L3-5	0.00 $\pm$ 0.00	-3.04 $\pm$ 1.77	-9.67 $\pm$ 3.64	-17.90 $\pm$ 4.52	-23.18 $\pm$ 4.48	-24.87 $\pm$ 5.22	-21.37 $\pm$ 5.27	-15.37 $\pm$ 5.62	-6.73 $\pm$ 5.81	-0.98 $\pm$ 4.67	0.34 $\pm$ 2.50
	Global T10-L5	0.00 $\pm$ 0.00	-2.39 $\pm$ 2.21	-9.28 $\pm$ 4.49	-19.63 $\pm$ 5.38	-26.95 $\pm$ 4.50	-30.33 $\pm$ 4.65	-27.40 $\pm$ 4.62	-19.56 $\pm$ 8.40	-1.16 $\pm$ 8.74	4.38 $\pm$ 3.57	2.07 $\pm$ 1.53

Normalize Mean Difference $\pm$ SD for Weightlift 5kg: Semi Squat (Begin cycle)

	Segment	Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	6.06 $\pm$ 2.99	20.60 $\pm$ 10.35	42.51 $\pm$ 17.90	65.07 $\pm$ 20.49	85.38 $\pm$ 20.82	96.62 $\pm$ 16.99	102.53 $\pm$ 15.17	104.67 $\pm$ 15.11	105.07 $\pm$ 15.96	104.52 $\pm$ 16.13
	T4-6	0.00 $\pm$ 0.00	5.67 $\pm$ 3.35	20.53 $\pm$ 11.56	44.33 $\pm$ 20.00	70.23 $\pm$ 24.53	93.67 $\pm$ 25.33	108.66 $\pm$ 22.66	117.04 $\pm$ 19.83	120.74 $\pm$ 19.44	121.76 $\pm$ 20.61	122.07 $\pm$ 21.57
	T6-8	0.00 $\pm$ 0.00	2.89 $\pm$ 4.59	14.62 $\pm$ 14.58	36.48 $\pm$ 25.60	62.27 $\pm$ 30.21	86.46 $\pm$ 31.78	103.47 $\pm$ 31.14	114.21 $\pm$ 30.50	118.92 $\pm$ 30.00	120.12 $\pm$ 30.03	120.42 $\pm$ 30.52
	T8-10	0.00 $\pm$ 0.00	-3.40 $\pm$ 4.05	-1.94 $\pm$ 16.32	14.85 $\pm$ 29.58	38.23 $\pm$ 37.00	62.38 $\pm$ 38.14	79.95 $\pm$ 37.97	90.55 $\pm$ 37.15	94.76 $\pm$ 36.71	96.03 $\pm$ 36.83	96.55 $\pm$ 37.36
	Global T1-10	0.00 $\pm$ 0.00	1.18 $\pm$ 2.61	8.99 $\pm$ 12.09	28.51 $\pm$ 22.56	50.75 $\pm$ 26.34	72.59 $\pm$ 26.87	86.54 $\pm$ 24.81	94.52 $\pm$ 23.59	97.70 $\pm$ 23.66	98.34 $\pm$ 23.95	98.49 $\pm$ 24.41
Lumbar	T10-L1	0.00 $\pm$ 0.00	-5.31 $\pm$ 3.19	-11.44 $\pm$ 10.31	-2.20 $\pm$ 23.10	20.81 $\pm$ 27.41	44.10 $\pm$ 27.55	60.80 $\pm$ 27.31	71.30 $\pm$ 26.81	76.09 $\pm$ 26.55	77.73 $\pm$ 26.91	78.21 $\pm$ 27.69
	L1-3	0.00 $\pm$ 0.00	-3.22 $\pm$ 4.18	-7.59 $\pm$ 10.00	0.94 $\pm$ 19.50	22.11 $\pm$ 22.68	43.04 $\pm$ 23.50	58.39 $\pm$ 23.51	68.02 $\pm$ 23.07	72.27 $\pm$ 22.79	73.52 $\pm$ 23.42	73.79 $\pm$ 24.60
	L3-5	0.00 $\pm$ 0.00	0.24 $\pm$ 2.20	4.24 $\pm$ 5.80	15.92 $\pm$ 12.24	31.96 $\pm$ 15.20	48.23 $\pm$ 17.51	60.44 $\pm$ 19.24	68.15 $\pm$ 20.55	71.41 $\pm$ 21.35	72.15 $\pm$ 22.51	71.76 $\pm$ 23.53
	Global T10-L5	0.00 $\pm$ 0.00	-2.85 $\pm$ 2.11	-4.25 $\pm$ 6.03	5.87 $\pm$ 14.25	25.61 $\pm$ 17.58	45.43 $\pm$ 18.61	59.88 $\pm$ 19.14	69.03 $\pm$ 19.49	73.21 $\pm$ 19.94	74.44 $\pm$ 21.07	74.49 $\pm$ 22.27

Normalize Mean Difference $\pm$ SD for Weightlift 5kg: Semi Squat (Return cycle)

	Segment	Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	113.96 $\pm$ 19.11	114.31 $\pm$ 18.85	113.60 $\pm$ 19.68	109.76 $\pm$ 20.70	96.81 $\pm$ 18.29	75.94 $\pm$ 17.18	52.81 $\pm$ 16.10	30.73 $\pm$ 12.43	14.38 $\pm$ 6.24	4.32 $\pm$ 2.13	0.00 $\pm$ 0.00
	T4-6	129.30 $\pm$ 24.98	129.38 $\pm$ 25.39	128.60 $\pm$ 25.54	122.54 $\pm$ 23.96	107.38 $\pm$ 22.59	83.72 $\pm$ 21.90	57.27 $\pm$ 20.47	31.34 $\pm$ 15.94	13.46 $\pm$ 9.15	3.71 $\pm$ 3.18	0.00 $\pm$ 0.00
	T6-8	118.55 $\pm$ 34.63	118.19 $\pm$ 34.93	115.89 $\pm$ 33.49	108.04 $\pm$ 31.14	92.30 $\pm$ 29.42	68.60 $\pm$ 29.57	42.03 $\pm$ 28.80	17.45 $\pm$ 23.64	3.33 $\pm$ 13.95	-0.33 $\pm$ 4.60	0.00 $\pm$ 0.00
	T8-10	93.31 $\pm$ 37.30	92.84 $\pm$ 36.93	90.84 $\pm$ 35.75	83.59 $\pm$ 34.10	67.72 $\pm$ 32.03	44.27 $\pm$ 31.54	18.85 $\pm$ 30.97	-1.02 $\pm$ 22.05	-7.59 $\pm$ 11.15	-3.99 $\pm$ 3.39	0.00 $\pm$ 0.00
	Global T1-10	100.75 $\pm$ 24.54	100.80 $\pm$ 24.71	99.79 $\pm$ 24.13	94.15 $\pm$ 22.77	80.11 $\pm$ 20.80	58.32 $\pm$ 20.25	34.47 $\pm$ 20.14	14.11 $\pm$ 14.84	2.80 $\pm$ 6.96	-0.02 $\pm$ 1.63	0.00 $\pm$ 0.00
Lumbar	T10-L1	77.57 $\pm$ 28.27	77.02 $\pm$ 28.30	74.62 $\pm$ 27.27	67.06 $\pm$ 25.66	52.47 $\pm$ 24.79	30.71 $\pm$ 25.64	6.28 $\pm$ 25.77	-11.31 $\pm$ 17.53	-11.55 $\pm$ 6.50	-4.43 $\pm$ 2.32	0.00 $\pm$ 0.00
	L1-3	75.56 $\pm$ 26.54	74.87 $\pm$ 26.30	72.89 $\pm$ 25.43	66.35 $\pm$ 24.25	52.90 $\pm$ 23.21	33.26 $\pm$ 23.01	12.06 $\pm$ 22.49	-4.10 $\pm$ 15.51	-6.32 $\pm$ 9.49	-2.62 $\pm$ 3.35	0.00 $\pm$ 0.00
	L3-5	71.49 $\pm$ 23.59	70.93 $\pm$ 23.91	69.62 $\pm$ 24.02	64.27 $\pm$ 22.80	53.16 $\pm$ 20.40	37.59 $\pm$ 17.55	21.57 $\pm$ 14.65	8.15 $\pm$ 11.02	1.19 $\pm$ 7.39	-0.24 $\pm$ 3.11	0.00 $\pm$ 0.00
	Global T10-L5	74.04 $\pm$ 22.75	73.33 $\pm$ 22.88	71.36 $\pm$ 22.33	64.91 $\pm$ 20.88	51.92 $\pm$ 19.15	33.14 $\pm$ 18.38	12.90 $\pm$ 17.34	-2.28 $\pm$ 11.65	-5.67 $\pm$ 5.44	-2.33 $\pm$ 2.05	0.00 $\pm$ 0.00

Normalize Segmental Difference $\pm$ SD for Weightlift 5kg: Semi Squat (Begin cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	6.06 $\pm$ 2.99	14.54 $\pm$ 7.57	21.91 $\pm$ 7.95	22.56 $\pm$ 7.64	20.31 $\pm$ 7.91	11.24 $\pm$ 8.73	5.91 $\pm$ 7.35	2.15 $\pm$ 4.83	0.40 $\pm$ 3.64	-0.55 $\pm$ 2.87
	T4-6	0.00 $\pm$ 0.00	5.67 $\pm$ 3.35	14.86 $\pm$ 8.41	23.80 $\pm$ 9.00	25.90 $\pm$ 6.90	23.44 $\pm$ 7.25	14.99 $\pm$ 9.27	8.39 $\pm$ 7.53	3.70 $\pm$ 5.38	1.02 $\pm$ 3.81	0.31 $\pm$ 2.73
	T6-8	0.00 $\pm$ 0.00	2.89 $\pm$ 4.59	11.73 $\pm$ 10.69	21.86 $\pm$ 11.96	25.79 $\pm$ 7.07	24.19 $\pm$ 7.33	17.01 $\pm$ 7.58	10.75 $\pm$ 6.74	4.71 $\pm$ 5.30	1.20 $\pm$ 4.39	0.30 $\pm$ 3.22
	T8-10	0.00 $\pm$ 0.00	-3.40 $\pm$ 4.05	1.46 $\pm$ 13.43	16.78 $\pm$ 17.32	23.38 $\pm$ 10.75	24.16 $\pm$ 6.88	17.57 $\pm$ 7.15	10.60 $\pm$ 6.58	4.21 $\pm$ 5.76	1.27 $\pm$ 4.92	0.52 $\pm$ 3.37
	Global T1-10	0.00 $\pm$ 0.00	1.18 $\pm$ 2.61	7.80 $\pm$ 9.86	19.52 $\pm$ 12.39	22.24 $\pm$ 8.34	21.84 $\pm$ 7.14	13.96 $\pm$ 7.45	7.98 $\pm$ 7.13	3.18 $\pm$ 4.65	0.64 $\pm$ 3.25	0.16 $\pm$ 2.28
Lumbar	T10-L1	0.00 $\pm$ 0.00	-5.31 $\pm$ 3.19	-6.12 $\pm$ 10.05	9.24 $\pm$ 17.76	23.01 $\pm$ 9.05	23.29 $\pm$ 6.57	16.70 $\pm$ 7.12	10.50 $\pm$ 6.34	4.79 $\pm$ 5.32	1.64 $\pm$ 4.83	0.48 $\pm$ 3.43
	L1-3	0.00 $\pm$ 0.00	-3.22 $\pm$ 4.18	-4.37 $\pm$ 7.02	8.53 $\pm$ 14.14	21.18 $\pm$ 6.46	20.92 $\pm$ 6.12	15.35 $\pm$ 6.69	9.63 $\pm$ 6.05	4.25 $\pm$ 5.20	1.25 $\pm$ 4.70	0.27 $\pm$ 3.18
	L3-5	0.00 $\pm$ 0.00	0.24 $\pm$ 2.20	4.00 $\pm$ 4.41	11.68 $\pm$ 7.61	16.04 $\pm$ 5.13	16.27 $\pm$ 5.93	12.21 $\pm$ 6.06	7.71 $\pm$ 5.57	3.27 $\pm$ 4.95	0.74 $\pm$ 4.20	-0.39 $\pm$ 2.48
	Global T10-L5	0.00 $\pm$ 0.00	-2.85 $\pm$ 2.11	-1.40 $\pm$ 6.44	10.12 $\pm$ 11.43	19.73 $\pm$ 6.77	19.83 $\pm$ 5.94	14.44 $\pm$ 6.46	9.16 $\pm$ 6.04	4.17 $\pm$ 5.08	1.24 $\pm$ 4.40	0.05 $\pm$ 2.80

Normalize Segmental Difference $\pm$ SD for Weightlift 5kg: Semi Squat (Return cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	0.34 $\pm$ 3.00	-0.70 $\pm$ 4.91	-3.84 $\pm$ 5.86	-12.95 $\pm$ 8.25	-20.87 $\pm$ 8.21	-23.13 $\pm$ 7.18	-22.08 $\pm$ 5.58	-16.34 $\pm$ 6.57	-10.06 $\pm$ 4.41	-4.32 $\pm$ 2.13
	T4-6	0.00 $\pm$ 0.00	0.08 $\pm$ 3.00	-0.78 $\pm$ 3.82	-6.07 $\pm$ 5.74	-15.16 $\pm$ 6.29	-23.66 $\pm$ 6.73	-26.44 $\pm$ 5.20	-25.93 $\pm$ 5.75	-17.88 $\pm$ 7.51	-9.75 $\pm$ 6.15	-3.71 $\pm$ 3.18
	T6-8	0.00 $\pm$ 0.00	-0.36 $\pm$ 3.29	-2.31 $\pm$ 4.24	-7.84 $\pm$ 5.61	-15.74 $\pm$ 6.35	-23.71 $\pm$ 6.76	-26.57 $\pm$ 4.92	-24.58 $\pm$ 7.56	-14.12 $\pm$ 11.93	-3.67 $\pm$ 9.94	0.33 $\pm$ 4.60
	T8-10	0.00 $\pm$ 0.00	-0.47 $\pm$ 2.96	-2.00 $\pm$ 4.46	-7.25 $\pm$ 6.43	-15.87 $\pm$ 5.92	-23.45 $\pm$ 6.11	-25.42 $\pm$ 5.32	-19.87 $\pm$ 14.23	-6.57 $\pm$ 15.19	3.60 $\pm$ 9.27	3.99 $\pm$ 3.39
	Global T1-10	0.00 $\pm$ 0.00	0.04 $\pm$ 2.29	-1.00 $\pm$ 3.43	-5.65 $\pm$ 5.11	-14.04 $\pm$ 6.70	-21.79 $\pm$ 7.02	-23.85 $\pm$ 6.43	-20.36 $\pm$ 8.79	-11.31 $\pm$ 9.92	-2.81 $\pm$ 5.88	0.02 $\pm$ 1.63
Lumbar	T10-L1	0.00 $\pm$ 0.00	-0.54 $\pm$ 3.08	-2.40 $\pm$ 3.91	-7.55 $\pm$ 5.52	-14.59 $\pm$ 5.92	-21.76 $\pm$ 6.42	-24.43 $\pm$ 4.51	-17.60 $\pm$ 11.24	-0.23 $\pm$ 14.68	7.12 $\pm$ 5.44	4.43 $\pm$ 2.32
	L1-3	0.00 $\pm$ 0.00	-0.68 $\pm$ 2.63	-1.99 $\pm$ 3.73	-6.54 $\pm$ 5.18	-13.46 $\pm$ 5.45	-19.63 $\pm$ 5.67	-21.20 $\pm$ 3.59	-16.16 $\pm$ 10.95	-2.22 $\pm$ 11.20	3.70 $\pm$ 6.52	2.62 $\pm$ 3.35
	L3-5	0.00 $\pm$ 0.00	-0.56 $\pm$ 2.28	-1.31 $\pm$ 3.45	-5.35 $\pm$ 4.92	-11.11 $\pm$ 5.16	-15.57 $\pm$ 5.56	-16.02 $\pm$ 4.61	-13.42 $\pm$ 5.26	-6.97 $\pm$ 5.83	-1.42 $\pm$ 4.56	0.24 $\pm$ 3.11
	Global T10-L5	0.00 $\pm$ 0.00	-0.71 $\pm$ 2.49	-1.97 $\pm$ 3.71	-6.45 $\pm$ 5.21	-12.99 $\pm$ 5.43	-18.78 $\pm$ 5.67	-20.24 $\pm$ 4.03	-15.18 $\pm$ 7.87	-3.39 $\pm$ 10.04	3.34 $\pm$ 3.85	2.33 $\pm$ 2.05

Normalize Mean Difference $\pm$ SD for Weightlift 5kg: Squat (Begin cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	4.32 $\pm$ 3.47	14.30 $\pm$ 8.67	28.81 $\pm$ 14.80	42.13 $\pm$ 20.19	50.42 $\pm$ 23.81	53.25 $\pm$ 24.82	52.35 $\pm$ 24.09	49.61 $\pm$ 22.98	46.41 $\pm$ 21.61	44.16 $\pm$ 20.63
	T4-6	0.00 $\pm$ 0.00	4.42 $\pm$ 4.19	15.71 $\pm$ 10.37	31.90 $\pm$ 17.91	46.87 $\pm$ 23.64	57.51 $\pm$ 27.34	62.51 $\pm$ 28.65	62.98 $\pm$ 28.67	61.00 $\pm$ 28.20	57.84 $\pm$ 27.13	55.40 $\pm$ 25.94
	T6-8	0.00 $\pm$ 0.00	1.87 $\pm$ 4.14	9.15 $\pm$ 11.80	23.76 $\pm$ 20.54	39.11 $\pm$ 26.52	50.54 $\pm$ 29.99	56.35 $\pm$ 31.75	57.57 $\pm$ 32.72	55.95 $\pm$ 33.11	52.84 $\pm$ 32.34	50.33 $\pm$ 31.47
	T8-10	0.00 $\pm$ 0.00	-3.16 $\pm$ 4.34	-5.21 $\pm$ 12.64	4.01 $\pm$ 21.74	17.29 $\pm$ 28.65	28.52 $\pm$ 31.64	34.59 $\pm$ 32.46	36.20 $\pm$ 32.68	34.99 $\pm$ 32.53	32.22 $\pm$ 31.43	30.10 $\pm$ 30.51
	Global T1-10	0.00 $\pm$ 0.00	0.54 $\pm$ 2.73	4.26 $\pm$ 9.17	16.11 $\pm$ 16.80	29.06 $\pm$ 22.53	38.48 $\pm$ 25.63	42.87 $\pm$ 26.02	43.02 $\pm$ 25.46	41.02 $\pm$ 24.82	38.07 $\pm$ 23.76	35.93 $\pm$ 22.86
Lumbar	T10-L1	0.00 $\pm$ 0.00	-4.18 $\pm$ 3.26	-8.36 $\pm$ 8.60	-5.84 $\pm$ 21.63	4.98 $\pm$ 28.35	15.35 $\pm$ 31.11	21.71 $\pm$ 29.74	23.55 $\pm$ 27.90	22.58 $\pm$ 26.70	19.90 $\pm$ 25.24	17.76 $\pm$ 23.82
	L1-3	0.00 $\pm$ 0.00	-3.18 $\pm$ 4.74	-4.84 $\pm$ 8.98	1.45 $\pm$ 18.14	12.24 $\pm$ 23.95	22.86 $\pm$ 24.39	28.66 $\pm$ 22.72	29.84 $\pm$ 21.29	28.16 $\pm$ 21.06	24.94 $\pm$ 20.97	22.66 $\pm$ 20.65
	L3-5	0.00 $\pm$ 0.00	0.75 $\pm$ 2.53	5.22 $\pm$ 6.63	16.06 $\pm$ 10.00	27.40 $\pm$ 12.26	35.65 $\pm$ 13.01	38.71 $\pm$ 12.85	37.24 $\pm$ 13.06	33.48 $\pm$ 14.07	28.79 $\pm$ 14.94	25.80 $\pm$ 15.39
	Global T10-L5	0.00 $\pm$ 0.00	-2.08 $\pm$ 2.28	-2.14 $\pm$ 6.28	4.47 $\pm$ 14.27	15.19 $\pm$ 19.31	24.22 $\pm$ 21.32	28.83 $\pm$ 20.41	29.01 $\pm$ 19.50	26.60 $\pm$ 19.29	22.93 $\pm$ 18.55	20.34 $\pm$ 17.99

Normalize Mean Difference $\pm$ SD for Weightlift 5kg: Squat (Return cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	53.32 $\pm$ 21.08	55.11 $\pm$ 20.38	58.04 $\pm$ 19.05	58.17 $\pm$ 17.53	53.15 $\pm$ 16.28	43.60 $\pm$ 14.56	31.94 $\pm$ 11.97	19.73 $\pm$ 8.55	9.73 $\pm$ 4.84	2.94 $\pm$ 1.83	0.00 $\pm$ 0.00
	T4-6	60.15 $\pm$ 27.58	62.38 $\pm$ 26.67	65.75 $\pm$ 25.37	65.86 $\pm$ 24.06	59.85 $\pm$ 22.59	48.90 $\pm$ 20.66	35.92 $\pm$ 16.98	22.13 $\pm$ 12.29	10.92 $\pm$ 7.07	3.45 $\pm$ 2.57	0.00 $\pm$ 0.00
	T6-8	47.56 $\pm$ 32.79	49.37 $\pm$ 31.23	52.61 $\pm$ 29.18	52.46 $\pm$ 27.59	46.04 $\pm$ 26.31	34.76 $\pm$ 25.13	21.88 $\pm$ 23.29	9.25 $\pm$ 18.73	2.15 $\pm$ 10.31	-0.48 $\pm$ 3.18	0.00 $\pm$ 0.00
	T8-10	28.18 $\pm$ 31.11	29.88 $\pm$ 30.06	32.67 $\pm$ 28.30	32.59 $\pm$ 26.87	26.61 $\pm$ 26.38	16.14 $\pm$ 24.90	3.92 $\pm$ 22.63	-5.25 $\pm$ 17.53	-6.39 $\pm$ 10.06	-2.91 $\pm$ 3.56	0.00 $\pm$ 0.00
	Global T1-10	39.66 $\pm$ 22.87	41.37 $\pm$ 21.82	44.24 $\pm$ 19.92	44.36 $\pm$ 18.70	39.05 $\pm$ 18.51	29.45 $\pm$ 17.59	17.86 $\pm$ 15.93	7.40 $\pm$ 12.11	1.65 $\pm$ 6.76	0.10 $\pm$ 2.41	0.00 $\pm$ 0.00
Lumbar	T10-L1	19.93 $\pm$ 26.77	21.76 $\pm$ 25.65	25.29 $\pm$ 23.63	25.76 $\pm$ 22.53	20.00 $\pm$ 22.97	9.35 $\pm$ 23.97	-1.59 $\pm$ 22.36	-8.13 $\pm$ 15.81	-8.03 $\pm$ 8.71	-3.28 $\pm$ 2.70	0.00 $\pm$ 0.00
	L1-3	26.54 $\pm$ 24.64	28.95 $\pm$ 23.61	33.29 $\pm$ 21.52	34.81 $\pm$ 20.14	30.29 $\pm$ 20.14	20.77 $\pm$ 20.73	10.12 $\pm$ 19.11	1.03 $\pm$ 14.34	-3.22 $\pm$ 8.79	-1.92 $\pm$ 3.15	0.00 $\pm$ 0.00
	L3-5	24.93 $\pm$ 14.38	27.90 $\pm$ 14.12	33.73 $\pm$ 12.30	37.24 $\pm$ 10.78	35.25 $\pm$ 10.81	29.02 $\pm$ 10.76	21.00 $\pm$ 9.93	12.38 $\pm$ 7.78	5.60 $\pm$ 4.84	1.47 $\pm$ 2.19	0.00 $\pm$ 0.00
	Global T10-L5	20.87 $\pm$ 19.69	23.33 $\pm$ 18.57	27.98 $\pm$ 16.46	29.92 $\pm$ 15.47	26.11 $\pm$ 16.02	17.72 $\pm$ 16.74	8.23 $\pm$ 15.69	1.35 $\pm$ 10.82	-1.72 $\pm$ 6.02	-0.95 $\pm$ 2.22	0.00 $\pm$ 0.00

Normalize Segmental Difference $\pm$ SD for Weightlift 5kg: Squat (Begin cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	4.32 $\pm$ 3.47	9.98 $\pm$ 5.53	14.51 $\pm$ 6.90	13.31 $\pm$ 7.17	8.30 $\pm$ 6.68	2.82 $\pm$ 4.66	-0.89 $\pm$ 4.46	-2.74 $\pm$ 4.45	-3.20 $\pm$ 4.10	-2.25 $\pm$ 3.29
	T4-6	0.00 $\pm$ 0.00	4.42 $\pm$ 4.19	11.29 $\pm$ 6.72	16.19 $\pm$ 8.42	14.97 $\pm$ 7.69	10.64 $\pm$ 7.20	5.00 $\pm$ 6.06	0.47 $\pm$ 5.35	-1.98 $\pm$ 4.85	-3.15 $\pm$ 4.31	-2.44 $\pm$ 3.34
	T6-8	0.00 $\pm$ 0.00	1.87 $\pm$ 4.14	7.28 $\pm$ 8.13	14.61 $\pm$ 10.02	15.35 $\pm$ 7.86	11.43 $\pm$ 6.82	5.81 $\pm$ 6.35	1.22 $\pm$ 6.14	-1.62 $\pm$ 5.47	-3.10 $\pm$ 4.56	-2.51 $\pm$ 3.45
	T8-10	0.00 $\pm$ 0.00	-3.16 $\pm$ 4.34	-2.06 $\pm$ 10.27	9.23 $\pm$ 13.83	13.28 $\pm$ 10.90	11.22 $\pm$ 7.57	6.08 $\pm$ 6.81	1.60 $\pm$ 6.60	-1.20 $\pm$ 5.70	-2.78 $\pm$ 4.88	-2.12 $\pm$ 3.47
	Global T1-10	0.00 $\pm$ 0.00	0.54 $\pm$ 2.73	3.72 $\pm$ 7.07	11.85 $\pm$ 9.76	12.95 $\pm$ 8.29	9.42 $\pm$ 6.74	4.38 $\pm$ 5.19	0.16 $\pm$ 5.15	-2.01 $\pm$ 4.90	-2.94 $\pm$ 4.28	-2.14 $\pm$ 3.24
Lumbar	T10-L1	0.00 $\pm$ 0.00	-4.18 $\pm$ 3.26	-4.18 $\pm$ 9.13	2.52 $\pm$ 16.55	10.82 $\pm$ 11.44	10.37 $\pm$ 8.03	6.36 $\pm$ 6.68	1.84 $\pm$ 6.63	-0.97 $\pm$ 5.45	-2.68 $\pm$ 4.42	-2.15 $\pm$ 3.32
	L1-3	0.00 $\pm$ 0.00	-3.18 $\pm$ 4.74	-1.66 $\pm$ 6.86	6.29 $\pm$ 13.46	10.80 $\pm$ 9.13	10.61 $\pm$ 5.33	5.80 $\pm$ 6.03	1.18 $\pm$ 6.23	-1.69 $\pm$ 5.43	-3.21 $\pm$ 4.66	-2.29 $\pm$ 3.37
	L3-5	0.00 $\pm$ 0.00	0.75 $\pm$ 2.53	4.48 $\pm$ 4.97	10.84 $\pm$ 5.60	11.34 $\pm$ 5.01	8.25 $\pm$ 4.70	3.06 $\pm$ 5.32	-1.47 $\pm$ 5.84	-3.76 $\pm$ 5.81	-4.69 $\pm$ 4.98	-2.99 $\pm$ 3.12
	Global T10-L5	0.00 $\pm$ 0.00	-2.08 $\pm$ 2.28	-0.06 $\pm$ 6.74	6.61 $\pm$ 10.85	10.72 $\pm$ 8.07	9.04 $\pm$ 6.13	4.60 $\pm$ 5.82	0.19 $\pm$ 5.95	-2.41 $\pm$ 5.36	-3.67 $\pm$ 4.58	-2.59 $\pm$ 3.09

Normalize Segmental Difference $\pm$ SD for Weightlift 5kg: Squat (Return cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	1.78 $\pm$ 3.68	2.94 $\pm$ 4.46	0.12 $\pm$ 5.17	-5.01 $\pm$ 5.47	-9.55 $\pm$ 4.52	-11.66 $\pm$ 4.44	-12.21 $\pm$ 4.24	-9.99 $\pm$ 4.07	-6.80 $\pm$ 3.29	-2.94 $\pm$ 1.83
	T4-6	0.00 $\pm$ 0.00	2.22 $\pm$ 3.56	3.38 $\pm$ 4.48	0.11 $\pm$ 5.18	-6.00 $\pm$ 5.51	-10.96 $\pm$ 5.06	-12.97 $\pm$ 5.41	-13.79 $\pm$ 5.54	-11.21 $\pm$ 5.65	-7.47 $\pm$ 4.75	-3.45 $\pm$ 2.57
	T6-8	0.00 $\pm$ 0.00	1.80 $\pm$ 3.46	3.24 $\pm$ 5.10	-0.14 $\pm$ 5.57	-6.43 $\pm$ 5.62	-11.28 $\pm$ 5.04	-12.88 $\pm$ 5.23	-12.64 $\pm$ 6.79	-7.10 $\pm$ 9.36	-2.63 $\pm$ 7.93	0.48 $\pm$ 3.18
	T8-10	0.00 $\pm$ 0.00	1.70 $\pm$ 3.70	2.80 $\pm$ 4.87	-0.09 $\pm$ 5.60	-5.97 $\pm$ 5.83	-10.47 $\pm$ 6.00	-12.22 $\pm$ 7.40	-9.18 $\pm$ 9.78	-1.14 $\pm$ 11.18	3.48 $\pm$ 6.94	2.91 $\pm$ 3.56
	Global T1-10	0.00 $\pm$ 0.00	1.71 $\pm$ 3.47	2.87 $\pm$ 4.45	0.12 $\pm$ 5.15	-5.31 $\pm$ 5.39	-9.60 $\pm$ 5.11	-11.58 $\pm$ 5.21	-10.47 $\pm$ 6.20	-5.74 $\pm$ 7.03	-1.55 $\pm$ 4.55	-0.10 $\pm$ 2.41
Lumbar	T10-L1	0.00 $\pm$ 0.00	1.83 $\pm$ 3.31	3.53 $\pm$ 4.67	0.47 $\pm$ 5.26	-5.77 $\pm$ 5.40	-10.65 $\pm$ 4.86	-10.93 $\pm$ 6.27	-6.54 $\pm$ 10.18	0.09 $\pm$ 10.09	4.76 $\pm$ 6.57	3.28 $\pm$ 2.70
	L1-3	0.00 $\pm$ 0.00	2.41 $\pm$ 2.89	4.35 $\pm$ 4.15	1.51 $\pm$ 4.77	-4.52 $\pm$ 4.59	-9.52 $\pm$ 3.90	-10.65 $\pm$ 4.11	-9.09 $\pm$ 7.35	-4.25 $\pm$ 7.97	1.30 $\pm$ 6.23	1.92 $\pm$ 3.15
	L3-5	0.00 $\pm$ 0.00	2.98 $\pm$ 2.52	5.83 $\pm$ 3.87	3.51 $\pm$ 4.44	-1.98 $\pm$ 3.70	-6.23 $\pm$ 2.85	-8.02 $\pm$ 2.40	-8.62 $\pm$ 3.01	-6.78 $\pm$ 3.94	-4.13 $\pm$ 3.21	-1.47 $\pm$ 2.19
	Global T10-L5	0.00 $\pm$ 0.00	2.46 $\pm$ 2.89	4.65 $\pm$ 4.17	1.94 $\pm$ 4.73	-3.82 $\pm$ 4.32	-8.38 $\pm$ 3.57	-9.49 $\pm$ 4.23	-6.89 $\pm$ 6.96	-3.07 $\pm$ 6.99	0.77 $\pm$ 4.24	0.95 $\pm$ 2.22

Appendix L Normalize Segmental and Mean Differences during Weightlift of 10kg load using different approaches.

Normalize Mean Difference $\pm$ SD for Weightlift 10kg: Stoop (Begin cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	6.40 $\pm$ 4.08	22.46 $\pm$ 11.55	49.59 $\pm$ 19.97	79.46 $\pm$ 22.99	100.15 $\pm$ 20.19	106.16 $\pm$ 23.81	99.12 $\pm$ 26.16	90.26 $\pm$ 24.18	84.08 $\pm$ 23.36	82.09 $\pm$ 23.58
	T4-6	0.00 $\pm$ 0.00	5.92 $\pm$ 4.60	22.45 $\pm$ 13.63	51.60 $\pm$ 25.43	86.99 $\pm$ 30.84	117.78 $\pm$ 24.95	130.22 $\pm$ 15.99	131.29 $\pm$ 22.33	125.39 $\pm$ 26.06	118.57 $\pm$ 26.77	116.07 $\pm$ 27.36
	T6-8	0.00 $\pm$ 0.00	2.76 $\pm$ 6.02	15.63 $\pm$ 16.49	43.15 $\pm$ 28.41	79.37 $\pm$ 33.91	115.01 $\pm$ 32.21	138.57 $\pm$ 21.53	149.71 $\pm$ 12.32	150.52 $\pm$ 13.49	146.50 $\pm$ 14.76	144.37 $\pm$ 14.98
	T8-10	0.00 $\pm$ 0.00	-3.22 $\pm$ 5.68	-1.73 $\pm$ 16.59	18.34 $\pm$ 32.58	52.90 $\pm$ 38.72	89.27 $\pm$ 36.59	118.60 $\pm$ 29.83	137.96 $\pm$ 20.27	145.80 $\pm$ 14.09	147.93 $\pm$ 12.49	146.93 $\pm$ 12.82
	Global T1-10	0.00 $\pm$ 0.00	1.37 $\pm$ 3.95	9.70 $\pm$ 12.94	33.32 $\pm$ 24.33	65.06 $\pm$ 27.16	92.03 $\pm$ 18.83	109.96 $\pm$ 11.71	115.63 $\pm$ 13.07	114.79 $\pm$ 13.77	112.31 $\pm$ 13.07	110.73 $\pm$ 14.01
Lumbar	T10-L1	0.00 $\pm$ 0.00	-4.25 $\pm$ 3.74	-8.58 $\pm$ 12.27	3.24 $\pm$ 29.28	35.05 $\pm$ 35.05	69.98 $\pm$ 32.81	99.00 $\pm$ 27.11	120.30 $\pm$ 21.02	132.53 $\pm$ 17.07	137.74 $\pm$ 15.69	138.62 $\pm$ 15.72
	L1-3	0.00 $\pm$ 0.00	-2.87 $\pm$ 3.93	-5.19 $\pm$ 10.30	5.26 $\pm$ 24.45	33.10 $\pm$ 31.02	64.90 $\pm$ 30.71	91.41 $\pm$ 27.21	110.90 $\pm$ 23.59	122.75 $\pm$ 22.77	128.41 $\pm$ 22.37	130.05 $\pm$ 22.48
	L3-5	0.00 $\pm$ 0.00	0.90 $\pm$ 2.03	5.32 $\pm$ 6.45	19.40 $\pm$ 11.77	41.81 $\pm$ 16.48	66.86 $\pm$ 18.93	88.48 $\pm$ 19.47	104.90 $\pm$ 20.45	115.17 $\pm$ 22.50	120.72 $\pm$ 24.46	122.45 $\pm$ 25.57
	Global T10-L5	0.00 $\pm$ 0.00	-1.98 $\pm$ 2.44	-1.98 $\pm$ 8.23	10.76 $\pm$ 18.80	37.87 $\pm$ 23.63	68.07 $\pm$ 22.87	93.56 $\pm$ 18.94	112.52 $\pm$ 15.43	124.21 $\pm$ 15.83	130.00 $\pm$ 17.02	131.54 $\pm$ 17.82

Normalize Mean Difference $\pm$ SD for Weightlift 10kg: Stoop (Return cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	91.11 $\pm$ 29.75	94.08 $\pm$ 31.14	100.49 $\pm$ 33.73	110.63 $\pm$ 32.90	111.18 $\pm$ 31.78	88.45 $\pm$ 28.97	57.98 $\pm$ 24.15	31.49 $\pm$ 15.68	13.96 $\pm$ 8.48	4.16 $\pm$ 3.61	0.00 $\pm$ 0.00
	T4-6	119.56 $\pm$ 33.60	122.19 $\pm$ 33.77	129.62 $\pm$ 32.97	137.65 $\pm$ 31.59	127.63 $\pm$ 30.69	96.38 $\pm$ 30.00	61.07 $\pm$ 24.57	31.16 $\pm$ 17.40	11.11 $\pm$ 11.01	2.34 $\pm$ 3.84	0.00 $\pm$ 0.00
	T6-8	134.01 $\pm$ 29.89	135.32 $\pm$ 30.02	139.08 $\pm$ 28.99	136.61 $\pm$ 30.64	114.07 $\pm$ 35.04	78.17 $\pm$ 34.55	41.99 $\pm$ 29.92	12.79 $\pm$ 23.21	-1.47 $\pm$ 13.57	-1.65 $\pm$ 4.33	0.00 $\pm$ 0.00
	T8-10	133.62 $\pm$ 29.88	133.50 $\pm$ 30.07	130.93 $\pm$ 30.18	115.39 $\pm$ 31.08	88.24 $\pm$ 33.23	52.53 $\pm$ 33.56	17.26 $\pm$ 31.44	-8.38 $\pm$ 26.02	-9.60 $\pm$ 13.23	-3.05 $\pm$ 4.51	0.00 $\pm$ 0.00
	Global T1-10	107.80 $\pm$ 24.17	109.83 $\pm$ 24.60	112.50 $\pm$ 24.38	108.79 $\pm$ 22.95	96.89 $\pm$ 24.73	68.19 $\pm$ 26.24	36.20 $\pm$ 23.60	10.45 $\pm$ 16.86	1.72 $\pm$ 7.62	0.47 $\pm$ 2.52	0.00 $\pm$ 0.00
Lumbar	T10-L1	130.73 $\pm$ 30.18	129.00 $\pm$ 30.23	120.84 $\pm$ 29.41	103.21 $\pm$ 29.79	75.48 $\pm$ 30.28	41.77 $\pm$ 31.03	9.78 $\pm$ 28.63	-11.10 $\pm$ 21.40	-11.21 $\pm$ 10.77	-4.23 $\pm$ 3.17	0.00 $\pm$ 0.00
	L1-3	126.28 $\pm$ 30.79	124.01 $\pm$ 30.87	115.45 $\pm$ 31.10	98.25 $\pm$ 32.26	73.24 $\pm$ 32.34	43.41 $\pm$ 31.18	16.29 $\pm$ 25.43	-2.10 $\pm$ 17.04	-5.96 $\pm$ 9.50	-2.33 $\pm$ 3.39	0.00 $\pm$ 0.00
	L3-5	117.57 $\pm$ 32.20	114.76 $\pm$ 32.20	105.76 $\pm$ 32.08	89.43 $\pm$ 31.39	68.33 $\pm$ 29.66	44.88 $\pm$ 25.78	24.22 $\pm$ 20.13	9.60 $\pm$ 13.62	2.13 $\pm$ 7.75	0.10 $\pm$ 2.87	0.00 $\pm$ 0.00
	Global T10-L5	124.52 $\pm$ 28.25	122.09 $\pm$ 28.04	113.00 $\pm$ 27.24	95.63 $\pm$ 27.50	71.05 $\pm$ 26.90	42.16 $\pm$ 25.24	15.31 $\pm$ 21.75	-2.23 $\pm$ 14.81	-4.83 $\pm$ 6.81	-2.08 $\pm$ 2.22	0.00 $\pm$ 0.00

Normalize Segmental Difference $\pm$ SD for Weightlift 10kg: Stoop (Begin cycle)

	Segment	Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	6.40 $\pm$ 4.08	16.05 $\pm$ 7.86	27.13 $\pm$ 9.76	29.87 $\pm$ 10.11	20.69 $\pm$ 21.54	6.02 $\pm$ 19.95	-7.04 $\pm$ 12.57	-8.85 $\pm$ 8.19	-6.18 $\pm$ 6.13	-1.99 $\pm$ 4.10
	T4-6	0.00 $\pm$ 0.00	5.92 $\pm$ 4.60	16.53 $\pm$ 9.55	29.15 $\pm$ 12.46	35.39 $\pm$ 8.07	30.78 $\pm$ 14.32	12.44 $\pm$ 23.51	1.07 $\pm$ 18.33	-5.91 $\pm$ 9.63	-6.82 $\pm$ 5.04	-2.50 $\pm$ 2.82
	T6-8	0.00 $\pm$ 0.00	2.76 $\pm$ 6.02	12.87 $\pm$ 11.39	27.52 $\pm$ 13.31	36.22 $\pm$ 7.70	35.64 $\pm$ 7.04	23.56 $\pm$ 16.33	11.14 $\pm$ 19.22	0.80 $\pm$ 11.15	-4.02 $\pm$ 6.26	-2.13 $\pm$ 2.58
	T8-10	0.00 $\pm$ 0.00	-3.22 $\pm$ 5.68	1.49 $\pm$ 13.33	20.07 $\pm$ 20.85	34.56 $\pm$ 8.99	36.37 $\pm$ 5.52	29.33 $\pm$ 9.22	19.35 $\pm$ 12.95	7.85 $\pm$ 10.54	2.12 $\pm$ 7.20	-1.00 $\pm$ 3.47
	Global T1-10	0.00 $\pm$ 0.00	1.37 $\pm$ 3.95	8.34 $\pm$ 10.31	23.62 $\pm$ 14.30	31.74 $\pm$ 8.34	26.97 $\pm$ 13.51	17.93 $\pm$ 13.27	5.67 $\pm$ 8.72	-0.84 $\pm$ 3.88	-2.48 $\pm$ 5.00	-1.58 $\pm$ 2.79
Lumbar	T10-L1	0.00 $\pm$ 0.00	-4.25 $\pm$ 3.74	-4.33 $\pm$ 10.71	11.81 $\pm$ 22.28	31.82 $\pm$ 9.16	34.93 $\pm$ 5.67	29.02 $\pm$ 8.31	21.30 $\pm$ 9.76	12.23 $\pm$ 8.27	5.21 $\pm$ 4.81	0.88 $\pm$ 3.45
	L1-3	0.00 $\pm$ 0.00	-2.87 $\pm$ 3.93	-2.33 $\pm$ 9.27	10.45 $\pm$ 18.46	27.84 $\pm$ 9.33	31.80 $\pm$ 5.36	26.51 $\pm$ 8.12	19.49 $\pm$ 9.84	11.85 $\pm$ 7.31	5.66 $\pm$ 4.19	1.64 $\pm$ 2.35
	L3-5	0.00 $\pm$ 0.00	0.90 $\pm$ 2.03	4.42 $\pm$ 4.80	14.08 $\pm$ 7.43	22.41 $\pm$ 7.09	25.06 $\pm$ 5.78	21.62 $\pm$ 6.62	16.42 $\pm$ 7.84	10.27 $\pm$ 5.96	5.54 $\pm$ 3.92	1.74 $\pm$ 1.87
	Global T10-L5	0.00 $\pm$ 0.00	-1.98 $\pm$ 2.44	0.00 $\pm$ 7.23	12.74 $\pm$ 13.96	27.11 $\pm$ 7.43	30.20 $\pm$ 5.11	25.49 $\pm$ 7.33	18.96 $\pm$ 9.08	11.69 $\pm$ 6.99	5.78 $\pm$ 3.93	1.54 $\pm$ 2.13

Normalize Segmental Difference $\pm$ SD for Weightlift 10kg: Stoop (Return cycle)

	Segment	Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	2.97 $\pm$ 6.22	6.41 $\pm$ 5.84	10.15 $\pm$ 15.45	0.55 $\pm$ 23.33	-22.73 $\pm$ 20.39	-30.47 $\pm$ 10.61	-26.49 $\pm$ 9.41	-17.53 $\pm$ 7.96	-9.79 $\pm$ 5.38	-4.16 $\pm$ 3.61
	T4-6	0.00 $\pm$ 0.00	2.63 $\pm$ 2.22	7.43 $\pm$ 5.70	8.03 $\pm$ 20.27	-10.01 $\pm$ 22.19	-31.25 $\pm$ 14.92	-35.31 $\pm$ 9.34	-29.91 $\pm$ 10.17	-20.05 $\pm$ 9.41	-8.77 $\pm$ 7.74	-2.34 $\pm$ 3.84
	T6-8	0.00 $\pm$ 0.00	1.32 $\pm$ 2.54	3.76 $\pm$ 9.55	-2.47 $\pm$ 16.81	-22.54 $\pm$ 17.50	-35.90 $\pm$ 9.96	-36.18 $\pm$ 9.44	-29.21 $\pm$ 11.41	-14.25 $\pm$ 13.59	-0.18 $\pm$ 10.50	1.65 $\pm$ 4.33
	T8-10	0.00 $\pm$ 0.00	-0.11 $\pm$ 3.02	-2.57 $\pm$ 9.97	-15.55 $\pm$ 14.35	-27.15 $\pm$ 12.11	-35.72 $\pm$ 9.42	-35.27 $\pm$ 9.39	-25.65 $\pm$ 14.68	-1.21 $\pm$ 17.48	6.55 $\pm$ 9.63	3.05 $\pm$ 4.51
	Global T1-10	0.00 $\pm$ 0.00	2.03 $\pm$ 2.85	2.67 $\pm$ 5.24	-3.71 $\pm$ 10.77	-11.90 $\pm$ 15.09	-28.70 $\pm$ 14.15	-31.99 $\pm$ 9.85	-25.75 $\pm$ 11.08	-8.73 $\pm$ 11.44	-1.24 $\pm$ 5.64	-0.47 $\pm$ 2.52
Lumbar	T10-L1	0.00 $\pm$ 0.00	-1.74 $\pm$ 3.08	-8.15 $\pm$ 7.45	-17.64 $\pm$ 10.62	-27.73 $\pm$ 9.20	-33.71 $\pm$ 8.85	-31.99 $\pm$ 9.66	-20.88 $\pm$ 16.26	-0.11 $\pm$ 15.80	6.98 $\pm$ 8.61	4.23 $\pm$ 3.17
	L1-3	0.00 $\pm$ 0.00	-2.27 $\pm$ 2.59	-8.55 $\pm$ 6.11	-17.20 $\pm$ 8.59	-25.01 $\pm$ 8.30	-29.84 $\pm$ 8.32	-27.12 $\pm$ 10.22	-18.39 $\pm$ 14.11	-3.86 $\pm$ 13.04	3.63 $\pm$ 7.08	2.33 $\pm$ 3.39
	L3-5	0.00 $\pm$ 0.00	-2.82 $\pm$ 2.33	-9.00 $\pm$ 5.36	-16.33 $\pm$ 7.13	-21.10 $\pm$ 7.04	-23.45 $\pm$ 7.43	-20.66 $\pm$ 8.29	-14.61 $\pm$ 9.24	-7.48 $\pm$ 7.84	-2.03 $\pm$ 5.79	-0.10 $\pm$ 2.87
	Global T10-L5	0.00 $\pm$ 0.00	-2.43 $\pm$ 2.26	-9.09 $\pm$ 5.95	-17.37 $\pm$ 8.87	-24.57 $\pm$ 8.02	-28.89 $\pm$ 7.95	-26.84 $\pm$ 8.24	-17.54 $\pm$ 12.65	-2.60 $\pm$ 10.95	2.74 $\pm$ 5.38	2.08 $\pm$ 2.22

Normalize Mean Difference $\pm$ SD for Weightlift 10kg: Semi Squat (Begin cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	6.22 $\pm$ 2.95	22.16 $\pm$ 9.32	46.66 $\pm$ 16.62	71.61 $\pm$ 21.10	89.17 $\pm$ 19.65	97.59 $\pm$ 17.16	101.05 $\pm$ 16.80	101.35 $\pm$ 16.39	101.35 $\pm$ 16.58	101.87 $\pm$ 16.79
	T4-6	0.00 $\pm$ 0.00	5.85 $\pm$ 3.57	22.24 $\pm$ 11.40	49.30 $\pm$ 19.09	77.02 $\pm$ 24.46	99.07 $\pm$ 25.70	112.27 $\pm$ 24.79	119.04 $\pm$ 24.05	121.24 $\pm$ 23.67	120.89 $\pm$ 22.95	119.68 $\pm$ 22.74
	T6-8	0.00 $\pm$ 0.00	3.23 $\pm$ 4.95	17.02 $\pm$ 13.22	42.73 $\pm$ 23.00	70.36 $\pm$ 30.19	93.94 $\pm$ 33.64	108.91 $\pm$ 33.94	116.43 $\pm$ 33.03	118.14 $\pm$ 31.84	117.51 $\pm$ 30.77	116.88 $\pm$ 30.18
	T8-10	0.00 $\pm$ 0.00	-3.66 $\pm$ 4.48	-0.18 $\pm$ 14.10	22.58 $\pm$ 25.19	49.30 $\pm$ 32.95	72.84 $\pm$ 36.98	88.84 $\pm$ 38.97	96.72 $\pm$ 38.74	98.79 $\pm$ 37.91	98.30 $\pm$ 36.54	97.06 $\pm$ 35.30
	Global T1-10	0.00 $\pm$ 0.00	0.95 $\pm$ 1.90	10.91 $\pm$ 8.74	34.17 $\pm$ 18.24	59.50 $\pm$ 24.50	80.16 $\pm$ 26.05	91.58 $\pm$ 25.42	96.87 $\pm$ 24.15	98.17 $\pm$ 23.47	97.94 $\pm$ 23.10	97.44 $\pm$ 22.75
Lumbar	T10-L1	0.00 $\pm$ 0.00	-4.84 $\pm$ 4.21	-11.11 $\pm$ 14.24	6.05 $\pm$ 22.74	31.98 $\pm$ 27.69	54.68 $\pm$ 30.44	69.39 $\pm$ 31.51	76.60 $\pm$ 31.11	79.11 $\pm$ 30.48	78.80 $\pm$ 29.38	77.25 $\pm$ 28.31
	L1-3	0.00 $\pm$ 0.00	-3.03 $\pm$ 4.30	-6.78 $\pm$ 14.74	7.90 $\pm$ 23.44	30.90 $\pm$ 28.06	51.79 $\pm$ 29.69	65.55 $\pm$ 29.74	72.63 $\pm$ 28.53	74.75 $\pm$ 27.64	73.92 $\pm$ 27.23	71.99 $\pm$ 26.74
	L3-5	0.00 $\pm$ 0.00	0.70 $\pm$ 2.75	6.06 $\pm$ 9.13	20.81 $\pm$ 16.09	39.34 $\pm$ 20.27	56.22 $\pm$ 23.47	67.19 $\pm$ 25.02	72.02 $\pm$ 25.35	72.71 $\pm$ 25.75	70.86 $\pm$ 25.12	68.71 $\pm$ 24.67
	Global T10-L5	0.00 $\pm$ 0.00	-2.08 $\pm$ 2.96	-2.92 $\pm$ 8.71	12.94 $\pm$ 15.29	35.32 $\pm$ 19.93	55.20 $\pm$ 23.28	68.14 $\pm$ 24.80	74.22 $\pm$ 25.00	75.72 $\pm$ 25.24	74.50 $\pm$ 24.62	72.76 $\pm$ 23.69

Normalize Mean Difference $\pm$ SD for Weightlift 10kg: Semi Squat (Return cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	112.44 $\pm$ 20.19	112.76 $\pm$ 19.67	111.93 $\pm$ 19.46	106.71 $\pm$ 17.93	92.78 $\pm$ 16.01	72.68 $\pm$ 16.51	48.40 $\pm$ 12.44	26.56 $\pm$ 7.78	11.94 $\pm$ 4.53	3.51 $\pm$ 1.92	0.00 $\pm$ 0.00
	T4-6	126.76 $\pm$ 26.36	127.08 $\pm$ 25.85	127.00 $\pm$ 24.97	120.34 $\pm$ 22.77	103.95 $\pm$ 20.45	78.82 $\pm$ 17.98	51.34 $\pm$ 15.31	26.87 $\pm$ 11.86	11.57 $\pm$ 6.59	3.23 $\pm$ 2.52	0.00 $\pm$ 0.00
	T6-8	111.66 $\pm$ 31.34	111.91 $\pm$ 30.15	111.28 $\pm$ 29.06	103.60 $\pm$ 27.55	86.03 $\pm$ 26.08	60.44 $\pm$ 25.21	32.72 $\pm$ 24.24	9.45 $\pm$ 20.64	-0.24 $\pm$ 13.49	-1.40 $\pm$ 5.21	0.00 $\pm$ 0.00
	T8-10	87.68 $\pm$ 37.81	87.32 $\pm$ 36.66	86.25 $\pm$ 35.79	78.94 $\pm$ 34.70	61.87 $\pm$ 33.73	36.84 $\pm$ 33.53	10.39 $\pm$ 32.15	-7.45 $\pm$ 24.41	-7.43 $\pm$ 12.06	-3.18 $\pm$ 4.38	0.00 $\pm$ 0.00
	Global T1-10	97.09 $\pm$ 24.50	96.94 $\pm$ 23.89	96.55 $\pm$ 23.36	90.17 $\pm$ 21.45	75.14 $\pm$ 20.16	53.27 $\pm$ 21.22	28.24 $\pm$ 18.90	8.83 $\pm$ 14.15	1.72 $\pm$ 7.27	0.11 $\pm$ 2.68	0.00 $\pm$ 0.00
Lumbar	T10-L1	74.57 $\pm$ 28.74	74.37 $\pm$ 27.32	73.64 $\pm$ 25.80	66.44 $\pm$ 24.33	50.27 $\pm$ 23.86	26.83 $\pm$ 24.89	1.74 $\pm$ 25.92	-11.88 $\pm$ 17.25	-9.15 $\pm$ 7.66	-3.38 $\pm$ 2.50	0.00 $\pm$ 0.00
	L1-3	73.54 $\pm$ 27.61	72.86 $\pm$ 26.24	72.25 $\pm$ 24.79	65.75 $\pm$ 24.20	50.97 $\pm$ 24.19	30.02 $\pm$ 24.44	8.36 $\pm$ 23.91	-3.75 $\pm$ 15.49	-5.23 $\pm$ 7.56	-2.35 $\pm$ 2.69	0.00 $\pm$ 0.00
	L3-5	68.98 $\pm$ 25.43	68.79 $\pm$ 25.16	68.84 $\pm$ 24.54	63.51 $\pm$ 23.94	51.00 $\pm$ 22.34	34.40 $\pm$ 20.16	18.31 $\pm$ 18.12	7.25 $\pm$ 13.02	1.70 $\pm$ 7.37	-0.21 $\pm$ 3.27	0.00 $\pm$ 0.00
	Global T10-L5	70.62 $\pm$ 23.73	70.44 $\pm$ 22.53	70.20 $\pm$ 21.39	63.94 $\pm$ 20.28	49.32 $\pm$ 18.99	29.09 $\pm$ 18.69	8.41 $\pm$ 18.88	-3.43 $\pm$ 12.28	-4.56 $\pm$ 5.12	-2.26 $\pm$ 1.75	0.00 $\pm$ 0.00

Normalize Segmental Difference $\pm$ SD for Weightlift 10kg: Semi Squat (Begin cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	6.22 $\pm$ 2.95	15.94 $\pm$ 6.80	24.50 $\pm$ 7.90	24.95 $\pm$ 6.51	17.57 $\pm$ 7.69	8.41 $\pm$ 8.11	3.47 $\pm$ 6.76	0.30 $\pm$ 4.11	0.00 $\pm$ 4.11	0.53 $\pm$ 3.08
	T4-6	0.00 $\pm$ 0.00	5.85 $\pm$ 3.57	16.39 $\pm$ 8.13	27.05 $\pm$ 9.29	27.73 $\pm$ 7.80	22.04 $\pm$ 6.21	13.20 $\pm$ 7.32	6.77 $\pm$ 5.95	2.20 $\pm$ 4.67	-0.35 $\pm$ 3.83	-1.21 $\pm$ 3.86
	T6-8	0.00 $\pm$ 0.00	3.23 $\pm$ 4.95	13.80 $\pm$ 9.44	25.70 $\pm$ 11.65	27.63 $\pm$ 8.81	23.58 $\pm$ 7.12	14.98 $\pm$ 6.23	7.51 $\pm$ 5.84	1.71 $\pm$ 5.17	-0.62 $\pm$ 4.19	-0.63 $\pm$ 3.87
	T8-10	0.00 $\pm$ 0.00	-3.66 $\pm$ 4.48	3.48 $\pm$ 11.86	22.75 $\pm$ 16.63	26.73 $\pm$ 10.73	23.54 $\pm$ 7.57	16.00 $\pm$ 6.32	7.88 $\pm$ 5.60	2.07 $\pm$ 4.95	-0.49 $\pm$ 4.28	-1.24 $\pm$ 3.52
	Global T1-10	0.00 $\pm$ 0.00	0.95 $\pm$ 1.90	9.97 $\pm$ 7.99	23.26 $\pm$ 11.69	25.33 $\pm$ 8.15	20.66 $\pm$ 6.62	11.42 $\pm$ 7.08	5.29 $\pm$ 5.44	1.30 $\pm$ 4.17	-0.23 $\pm$ 3.19	-0.50 $\pm$ 2.35
Lumbar	T10-L1	0.00 $\pm$ 0.00	-4.84 $\pm$ 4.21	-6.27 $\pm$ 11.46	17.16 $\pm$ 16.00	25.93 $\pm$ 10.02	22.70 $\pm$ 7.05	14.71 $\pm$ 5.86	7.21 $\pm$ 5.54	2.51 $\pm$ 4.93	-0.32 $\pm$ 4.40	-1.55 $\pm$ 3.79
	L1-3	0.00 $\pm$ 0.00	-3.03 $\pm$ 4.30	-3.74 $\pm$ 11.27	14.67 $\pm$ 15.16	23.01 $\pm$ 8.75	20.89 $\pm$ 6.83	13.76 $\pm$ 5.96	7.08 $\pm$ 5.14	2.13 $\pm$ 4.24	-0.83 $\pm$ 3.66	-1.94 $\pm$ 3.08
	L3-5	0.00 $\pm$ 0.00	0.70 $\pm$ 2.75	5.36 $\pm$ 7.05	14.74 $\pm$ 8.79	18.53 $\pm$ 6.74	16.88 $\pm$ 7.14	10.98 $\pm$ 5.56	4.83 $\pm$ 4.36	0.69 $\pm$ 3.29	-1.85 $\pm$ 3.59	-2.15 $\pm$ 2.97
	Global T10-L5	0.00 $\pm$ 0.00	-2.08 $\pm$ 2.96	-0.84 $\pm$ 6.96	15.86 $\pm$ 11.13	22.38 $\pm$ 7.91	19.88 $\pm$ 6.87	12.94 $\pm$ 5.57	6.08 $\pm$ 4.90	1.50 $\pm$ 3.81	-1.22 $\pm$ 3.56	-1.74 $\pm$ 3.04

Normalize Segmental Difference $\pm$ SD for Weightlift 10kg: Semi Squat (Return cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	0.32 $\pm$ 3.14	-0.83 $\pm$ 4.12	-5.22 $\pm$ 5.27	-13.93 $\pm$ 10.02	-20.09 $\pm$ 9.45	-24.28 $\pm$ 6.09	-21.83 $\pm$ 5.85	-14.63 $\pm$ 3.94	-8.43 $\pm$ 3.20	-3.51 $\pm$ 1.92
	T4-6	0.00 $\pm$ 0.00	0.32 $\pm$ 4.11	-0.08 $\pm$ 6.00	-6.67 $\pm$ 7.16	-16.38 $\pm$ 8.42	-25.13 $\pm$ 6.78	-27.48 $\pm$ 6.57	-24.47 $\pm$ 5.56	-15.30 $\pm$ 5.89	-8.34 $\pm$ 4.75	-3.23 $\pm$ 2.52
	T6-8	0.00 $\pm$ 0.00	0.25 $\pm$ 4.52	-0.62 $\pm$ 6.69	-7.68 $\pm$ 7.46	-17.57 $\pm$ 8.14	-25.59 $\pm$ 6.98	-27.72 $\pm$ 6.51	-23.26 $\pm$ 8.13	-9.69 $\pm$ 9.98	-1.16 $\pm$ 8.98	1.40 $\pm$ 5.21
	T8-10	0.00 $\pm$ 0.00	-0.35 $\pm$ 4.23	-1.08 $\pm$ 6.59	-7.31 $\pm$ 7.21	-17.07 $\pm$ 8.09	-25.03 $\pm$ 6.51	-26.44 $\pm$ 7.38	-17.85 $\pm$ 13.53	0.02 $\pm$ 14.96	4.26 $\pm$ 8.39	3.18 $\pm$ 4.38
	Global T1-10	0.00 $\pm$ 0.00	-0.15 $\pm$ 3.28	-0.39 $\pm$ 4.55	-6.38 $\pm$ 7.02	-15.03 $\pm$ 9.04	-21.87 $\pm$ 8.12	-25.03 $\pm$ 6.81	-19.41 $\pm$ 8.04	-7.12 $\pm$ 8.38	-1.60 $\pm$ 4.90	-0.11 $\pm$ 2.68
Lumbar	T10-L1	0.00 $\pm$ 0.00	-0.20 $\pm$ 4.83	-0.73 $\pm$ 6.65	-7.21 $\pm$ 6.98	-16.16 $\pm$ 7.43	-23.44 $\pm$ 6.27	-25.10 $\pm$ 6.03	-13.61 $\pm$ 14.12	2.73 $\pm$ 12.26	5.77 $\pm$ 6.09	3.38 $\pm$ 2.50
	L1-3	0.00 $\pm$ 0.00	-0.69 $\pm$ 4.13	-0.61 $\pm$ 6.44	-6.50 $\pm$ 6.55	-14.78 $\pm$ 6.79	-20.95 $\pm$ 5.86	-21.65 $\pm$ 5.38	-12.12 $\pm$ 12.28	-1.48 $\pm$ 10.20	2.88 $\pm$ 5.54	2.35 $\pm$ 2.69
	L3-5	0.00 $\pm$ 0.00	-0.19 $\pm$ 3.41	0.05 $\pm$ 6.53	-5.32 $\pm$ 6.25	-12.52 $\pm$ 6.25	-16.60 $\pm$ 5.95	-16.09 $\pm$ 5.08	-11.06 $\pm$ 6.80	-5.55 $\pm$ 6.47	-1.91 $\pm$ 4.88	0.21 $\pm$ 3.27
	Global T10-L5	0.00 $\pm$ 0.00	-0.18 $\pm$ 3.93	-0.24 $\pm$ 6.65	-6.26 $\pm$ 6.64	-14.61 $\pm$ 6.71	-20.24 $\pm$ 5.97	-20.68 $\pm$ 5.00	-11.85 $\pm$ 9.33	-1.13 $\pm$ 8.28	2.30 $\pm$ 4.02	2.26 $\pm$ 1.75

Normalize Mean Difference $\pm$ SD for Weightlift 10kg: Squat (Begin cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	5.64 $\pm$ 3.86	16.93 $\pm$ 9.39	32.19 $\pm$ 15.41	46.06 $\pm$ 21.66	55.15 $\pm$ 26.30	58.33 $\pm$ 28.14	57.76 $\pm$ 28.09	55.33 $\pm$ 26.48	52.05 $\pm$ 23.86	49.41 $\pm$ 21.96
	T4-6	0.00 $\pm$ 0.00	5.75 $\pm$ 4.58	18.45 $\pm$ 11.71	35.42 $\pm$ 19.78	51.00 $\pm$ 26.79	62.43 $\pm$ 30.92	67.09 $\pm$ 32.38	66.80 $\pm$ 32.10	64.58 $\pm$ 30.95	61.40 $\pm$ 29.29	59.06 $\pm$ 28.15
	T6-8	0.00 $\pm$ 0.00	2.67 $\pm$ 5.50	11.04 $\pm$ 14.83	27.38 $\pm$ 22.53	43.85 $\pm$ 28.17	55.95 $\pm$ 32.30	61.20 $\pm$ 34.15	61.62 $\pm$ 34.33	59.74 $\pm$ 33.53	56.40 $\pm$ 32.33	53.97 $\pm$ 31.18
	T8-10	0.00 $\pm$ 0.00	-3.03 $\pm$ 6.39	-4.02 $\pm$ 15.64	6.20 $\pm$ 25.93	21.69 $\pm$ 31.44	33.86 $\pm$ 35.02	39.80 $\pm$ 36.09	40.80 $\pm$ 36.03	39.16 $\pm$ 35.31	35.74 $\pm$ 34.05	32.89 $\pm$ 32.98
	Global T1-10	0.00 $\pm$ 0.00	1.11 $\pm$ 4.04	5.92 $\pm$ 10.85	18.82 $\pm$ 18.55	33.26 $\pm$ 24.19	43.52 $\pm$ 28.11	48.02 $\pm$ 29.30	47.95 $\pm$ 28.98	45.90 $\pm$ 27.63	42.62 $\pm$ 25.83	39.95 $\pm$ 24.23
Lumbar	T10-L1	0.00 $\pm$ 0.00	-5.43 $\pm$ 3.62	-11.16 $\pm$ 12.40	-4.69 $\pm$ 22.18	9.30 $\pm$ 28.28	21.33 $\pm$ 30.21	26.98 $\pm$ 29.91	27.83 $\pm$ 28.58	26.33 $\pm$ 26.87	22.86 $\pm$ 25.22	20.05 $\pm$ 23.82
	L1-3	0.00 $\pm$ 0.00	-3.45 $\pm$ 5.62	-5.91 $\pm$ 11.32	2.34 $\pm$ 18.46	16.58 $\pm$ 20.55	27.81 $\pm$ 20.87	32.63 $\pm$ 20.48	32.59 $\pm$ 19.87	30.14 $\pm$ 19.81	26.05 $\pm$ 19.60	22.87 $\pm$ 19.60
	L3-5	0.00 $\pm$ 0.00	0.45 $\pm$ 3.37	6.24 $\pm$ 7.18	17.72 $\pm$ 11.11	29.31 $\pm$ 12.10	37.53 $\pm$ 13.55	39.86 $\pm$ 13.72	37.62 $\pm$ 13.26	33.24 $\pm$ 13.47	27.45 $\pm$ 13.46	23.26 $\pm$ 13.74
	Global T10-L5	0.00 $\pm$ 0.00	-2.71 $\pm$ 3.04	-2.43 $\pm$ 7.03	5.98 $\pm$ 14.87	18.31 $\pm$ 19.09	28.17 $\pm$ 21.19	32.28 $\pm$ 21.24	31.73 $\pm$ 20.23	28.75 $\pm$ 19.26	24.07 $\pm$ 18.13	20.63 $\pm$ 16.94

Normalize Mean Difference $\pm$ SD for Weightlift 10kg: Squat (Return cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	54.74 $\pm$ 21.29	56.90 $\pm$ 20.62	60.25 $\pm$ 20.11	61.21 $\pm$ 19.80	57.00 $\pm$ 19.47	47.04 $\pm$ 17.52	34.45 $\pm$ 14.40	20.95 $\pm$ 10.22	9.81 $\pm$ 5.93	2.92 $\pm$ 2.33	0.00 $\pm$ 0.00
	T4-6	61.55 $\pm$ 28.55	64.04 $\pm$ 26.85	68.61 $\pm$ 25.67	69.79 $\pm$ 24.31	63.82 $\pm$ 22.39	52.02 $\pm$ 20.07	37.76 $\pm$ 16.93	23.26 $\pm$ 11.82	11.12 $\pm$ 6.94	3.45 $\pm$ 2.79	0.00 $\pm$ 0.00
	T6-8	47.93 $\pm$ 30.21	49.67 $\pm$ 28.56	53.73 $\pm$ 27.28	54.91 $\pm$ 25.98	49.37 $\pm$ 24.46	37.38 $\pm$ 23.38	22.41 $\pm$ 21.68	7.50 $\pm$ 18.76	-0.02 $\pm$ 12.00	-1.08 $\pm$ 4.72	0.00 $\pm$ 0.00
	T8-10	27.79 $\pm$ 35.81	29.14 $\pm$ 34.81	32.47 $\pm$ 34.26	33.59 $\pm$ 33.50	28.38 $\pm$ 32.58	17.56 $\pm$ 30.77	4.59 $\pm$ 27.66	-6.53 $\pm$ 22.33	-6.76 $\pm$ 11.64	-2.65 $\pm$ 4.46	0.00 $\pm$ 0.00
	Global T1-10	39.93 $\pm$ 25.71	41.65 $\pm$ 24.63	45.10 $\pm$ 24.13	46.13 $\pm$ 23.57	41.60 $\pm$ 22.99	31.64 $\pm$ 21.25	19.33 $\pm$ 18.52	7.38 $\pm$ 13.88	1.78 $\pm$ 7.41	-0.03 $\pm$ 2.65	0.00 $\pm$ 0.00
Lumbar	T10-L1	19.10 $\pm$ 28.31	20.79 $\pm$ 27.08	24.67 $\pm$ 26.67	26.08 $\pm$ 26.76	21.25 $\pm$ 26.63	10.75 $\pm$ 26.13	-1.39 $\pm$ 24.04	-9.60 $\pm$ 18.06	-9.63 $\pm$ 11.11	-4.12 $\pm$ 4.86	0.00 $\pm$ 0.00
	L1-3	26.41 $\pm$ 21.03	28.25 $\pm$ 20.35	32.63 $\pm$ 19.22	34.82 $\pm$ 18.26	30.80 $\pm$ 17.89	21.27 $\pm$ 17.48	9.32 $\pm$ 16.84	-1.19 $\pm$ 13.54	-4.09 $\pm$ 7.60	-2.17 $\pm$ 2.67	0.00 $\pm$ 0.00
	L3-5	25.22 $\pm$ 11.87	27.62 $\pm$ 10.92	33.11 $\pm$ 10.63	36.84 $\pm$ 12.03	34.78 $\pm$ 13.49	28.10 $\pm$ 13.02	19.56 $\pm$ 11.15	10.61 $\pm$ 8.55	3.99 $\pm$ 5.87	0.60 $\pm$ 3.21	0.00 $\pm$ 0.00
	Global T10-L5	21.60 $\pm$ 17.89	23.59 $\pm$ 16.60	28.20 $\pm$ 16.07	30.72 $\pm$ 16.60	27.14 $\pm$ 17.44	18.58 $\pm$ 17.27	8.77 $\pm$ 14.68	0.34 $\pm$ 10.62	-2.72 $\pm$ 6.56	-1.31 $\pm$ 2.78	0.00 $\pm$ 0.00

Normalize Segmental Difference $\pm$ SD for Weightlift 10kg: Squat (Begin cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	5.64 $\pm$ 3.86	11.30 $\pm$ 6.03	15.26 $\pm$ 8.20	13.87 $\pm$ 9.31	9.09 $\pm$ 7.85	3.17 $\pm$ 4.44	-0.56 $\pm$ 4.17	-2.43 $\pm$ 4.76	-3.28 $\pm$ 5.54	-2.63 $\pm$ 4.85
	T4-6	0.00 $\pm$ 0.00	5.75 $\pm$ 4.58	12.70 $\pm$ 7.60	16.97 $\pm$ 9.67	15.59 $\pm$ 9.81	11.42 $\pm$ 7.58	4.66 $\pm$ 4.75	-0.29 $\pm$ 5.10	-2.22 $\pm$ 5.54	-3.18 $\pm$ 5.91	-2.35 $\pm$ 4.90
	T6-8	0.00 $\pm$ 0.00	2.67 $\pm$ 5.50	8.37 $\pm$ 9.83	16.34 $\pm$ 10.55	16.48 $\pm$ 9.61	12.10 $\pm$ 7.55	5.26 $\pm$ 4.90	0.42 $\pm$ 5.27	-1.88 $\pm$ 5.76	-3.35 $\pm$ 5.92	-2.42 $\pm$ 4.85
	T8-10	0.00 $\pm$ 0.00	-3.03 $\pm$ 6.39	-0.99 $\pm$ 10.84	10.22 $\pm$ 15.43	15.49 $\pm$ 10.85	12.17 $\pm$ 8.01	5.94 $\pm$ 5.24	1.01 $\pm$ 5.56	-1.64 $\pm$ 5.81	-3.43 $\pm$ 5.66	-2.84 $\pm$ 4.72
	Global T1-10	0.00 $\pm$ 0.00	1.11 $\pm$ 4.04	4.81 $\pm$ 7.46	12.90 $\pm$ 10.86	14.43 $\pm$ 9.57	10.26 $\pm$ 7.64	4.50 $\pm$ 4.53	-0.07 $\pm$ 4.63	-2.05 $\pm$ 5.09	-3.28 $\pm$ 5.27	-2.67 $\pm$ 4.64
Lumbar	T10-L1	0.00 $\pm$ 0.00	-5.43 $\pm$ 3.62	-5.74 $\pm$ 11.60	6.48 $\pm$ 16.26	13.99 $\pm$ 10.40	12.03 $\pm$ 7.20	5.65 $\pm$ 5.11	0.85 $\pm$ 5.36	-1.50 $\pm$ 5.45	-3.48 $\pm$ 5.12	-2.80 $\pm$ 4.24
	L1-3	0.00 $\pm$ 0.00	-3.45 $\pm$ 5.62	-2.47 $\pm$ 8.99	8.25 $\pm$ 12.80	14.24 $\pm$ 7.66	11.23 $\pm$ 5.85	4.82 $\pm$ 4.50	-0.04 $\pm$ 5.52	-2.45 $\pm$ 5.09	-4.09 $\pm$ 5.05	-3.18 $\pm$ 4.75
	L3-5	0.00 $\pm$ 0.00	0.45 $\pm$ 3.37	5.79 $\pm$ 6.02	11.48 $\pm$ 5.89	11.59 $\pm$ 6.05	8.23 $\pm$ 5.37	2.32 $\pm$ 4.30	-2.23 $\pm$ 5.25	-4.38 $\pm$ 5.23	-5.79 $\pm$ 4.80	-4.20 $\pm$ 3.80
	Global T10-L5	0.00 $\pm$ 0.00	-2.71 $\pm$ 3.04	0.27 $\pm$ 8.02	8.42 $\pm$ 11.01	12.33 $\pm$ 8.20	9.86 $\pm$ 6.34	4.10 $\pm$ 4.35	-0.55 $\pm$ 5.01	-2.97 $\pm$ 5.11	-4.68 $\pm$ 4.78	-3.44 $\pm$ 4.11

Normalize Segmental Difference $\pm$ SD for Weightlift 10kg: Squat (Return cycle)

Segment		Percentage of motion completion										
		0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Thoracic	T1-4	0.00 $\pm$ 0.00	2.16 $\pm$ 4.00	3.35 $\pm$ 5.11	0.96 $\pm$ 4.73	-4.21 $\pm$ 3.82	-9.96 $\pm$ 3.94	-12.59 $\pm$ 4.43	-13.50 $\pm$ 4.97	-11.14 $\pm$ 5.18	-6.88 $\pm$ 3.88	-2.92 $\pm$ 2.33
	T4-6	0.00 $\pm$ 0.00	2.49 $\pm$ 4.11	4.58 $\pm$ 5.94	1.18 $\pm$ 5.56	-5.97 $\pm$ 4.71	-11.80 $\pm$ 4.22	-14.26 $\pm$ 5.15	-14.49 $\pm$ 6.57	-12.14 $\pm$ 5.87	-7.68 $\pm$ 4.46	-3.45 $\pm$ 2.79
	T6-8	0.00 $\pm$ 0.00	1.75 $\pm$ 3.97	4.05 $\pm$ 6.04	1.18 $\pm$ 5.54	-5.53 $\pm$ 5.07	-11.99 $\pm$ 4.49	-14.97 $\pm$ 5.40	-14.91 $\pm$ 7.27	-7.53 $\pm$ 8.71	-1.05 $\pm$ 7.94	1.08 $\pm$ 4.72
	T8-10	0.00 $\pm$ 0.00	1.35 $\pm$ 4.13	3.33 $\pm$ 6.33	1.12 $\pm$ 5.48	-5.21 $\pm$ 5.11	-10.83 $\pm$ 5.53	-12.96 $\pm$ 8.13	-11.12 $\pm$ 10.98	-0.23 $\pm$ 12.97	4.11 $\pm$ 8.09	2.65 $\pm$ 4.46
	Global T1-10	0.00 $\pm$ 0.00	1.73 $\pm$ 3.81	3.44 $\pm$ 5.62	1.04 $\pm$ 4.97	-4.53 $\pm$ 4.27	-9.96 $\pm$ 4.56	-12.32 $\pm$ 5.63	-11.94 $\pm$ 7.54	-5.60 $\pm$ 7.64	-1.81 $\pm$ 5.32	0.03 $\pm$ 2.65
Lumbar	T10-L1	0.00 $\pm$ 0.00	1.69 $\pm$ 4.25	3.88 $\pm$ 6.02	1.42 $\pm$ 5.39	-4.84 $\pm$ 4.86	-10.50 $\pm$ 4.65	-12.14 $\pm$ 7.34	-8.21 $\pm$ 11.41	-0.02 $\pm$ 11.17	5.51 $\pm$ 6.64	4.12 $\pm$ 4.86
	L1-3	0.00 $\pm$ 0.00	1.83 $\pm$ 4.05	4.38 $\pm$ 5.57	2.19 $\pm$ 5.04	-4.02 $\pm$ 4.22	-9.52 $\pm$ 3.85	-11.96 $\pm$ 4.56	-10.51 $\pm$ 6.92	-2.89 $\pm$ 8.69	1.91 $\pm$ 5.80	2.17 $\pm$ 2.67
	L3-5	0.00 $\pm$ 0.00	2.40 $\pm$ 3.35	5.48 $\pm$ 5.11	3.73 $\pm$ 4.74	-2.07 $\pm$ 3.74	-6.68 $\pm$ 2.71	-8.54 $\pm$ 3.67	-8.95 $\pm$ 4.34	-6.63 $\pm$ 4.44	-3.38 $\pm$ 3.98	-0.60 $\pm$ 3.21
	Global T10-L5	0.00 $\pm$ 0.00	1.99 $\pm$ 3.63	4.62 $\pm$ 5.41	2.51 $\pm$ 4.97	-3.57 $\pm$ 4.20	-8.56 $\pm$ 3.32	-9.81 $\pm$ 5.93	-8.43 $\pm$ 7.46	-3.06 $\pm$ 7.25	1.41 $\pm$ 4.32	1.31 $\pm$ 2.78

Appendix M Simulation Analysis (Natrán-Patran and Solidworks) for initial concept design.

The most critical components of the proposed system that are susceptible to failure are the locking system (consist of gear gripper and rack), flat connector, and the location at which both bases are connected (on back base). These components bear the upper body's weight in the form of substantial load and are categorized as critical components due to the small dimension as shown in Figure A below. Thus, an initial stress analysis has been done to propose the most suitable design, dimension, and material for the proposed multi-joint spinal orthosis. For the simulation and finite element analysis (FEA), Solidworks Simulation 2017, and Patran-Nastran 2016 software have been utilized. The average Malaysian weight recorded is 58.3 ± 0.1 kg [252] and for analysis, the value is rounded to 60 kg. The spine only supports approximately half of the human total weight, thus only 30 kg will be used for the analysis and added with another 20 kg of load, considering the user is doing lifting works while wearing the orthosis. For the analysis, the material specified is Cast Alloy Steel for the Front base as it has reliable strength and more importantly, the lightweight characteristic compared to other steels. Also, few plastic group materials have been used for the analysis as they are significantly lighter than steel. For the rack, Alloy Steel is a suitable material as stronger steel is required to prevent the grip from failure during locking operation.

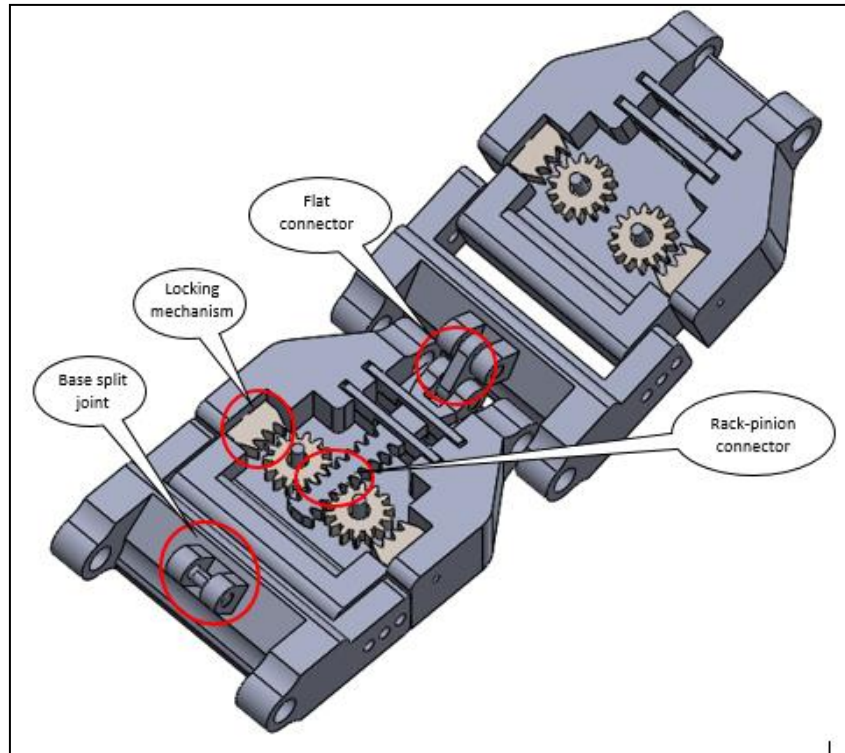


Figure 1 Components that are susceptible to failure when the load of 500N is applied during working operation.

a) Back base

During operation, the base will handle the compression force and in locking condition, it prevents excessive movement of the spine especially on the targeted vertebrae. As shown in Figure 2, the most critical region on the base is located at the geometry that connects between the upper and lower base. The other results obtained were shown in Figures 3 and 4.

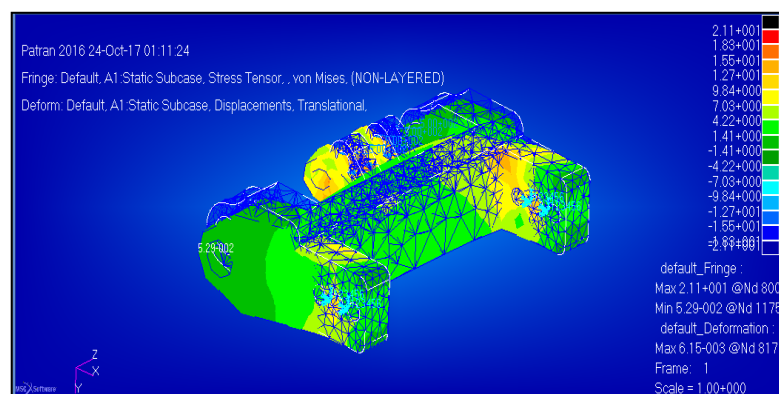


Figure 2 FEA analysis result for stress from Patran-Nastran for Back Base

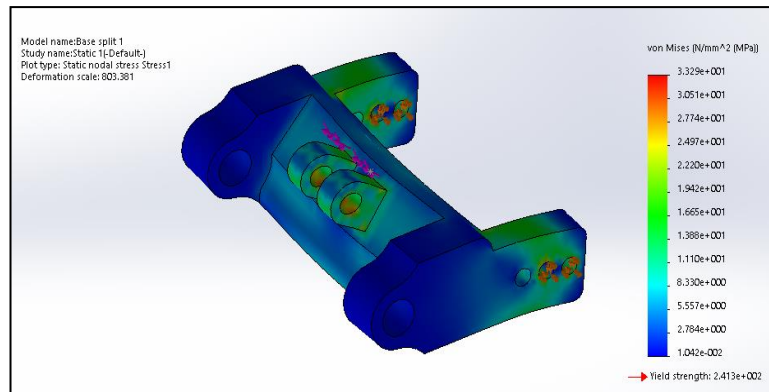


Figure 3 Stress result from Solidworks for Back base

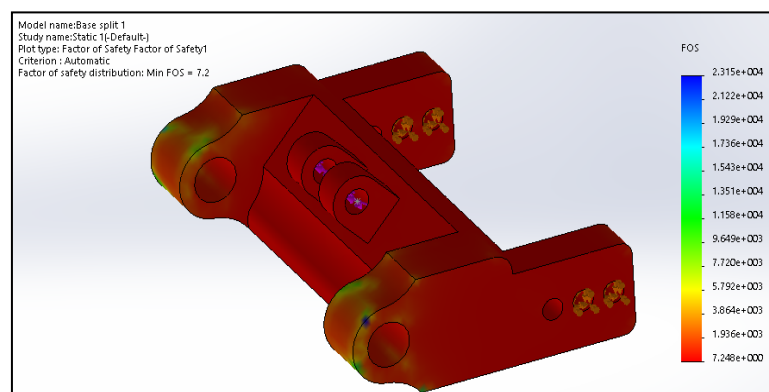


Figure 4 Factor of Safety result from Solidworks for Back Base

From the results obtained as shown in Table 1, Cast Alloy Steel used for the analysis can sustain 500N of load during spinal flexion and extension. The highest stress recorded is only 33.2 MPa compared to the yield strength of the material, which is 241 Mpa, but the material is quite higher in mass density as compared with the plastic. For the analysis of different materials of plastic, Acrylic Medium-High Impact shows better results compared with the others. Further improvement on the base has been conducted such as using hollow-shaped metal. Due to the big dimension compared to other components, this part should have a smaller mass density with enough strength to withstand the load, especially in the lumbar region.

Table 1 Summary of results of FEA for Back base

Base (Material)	FEA software used			
	Solidwork 2017		Patran Nastran 2016 Tet 10 (Nodes 7000, Element 4000)	
Cast steel alloy σ_Y : 241.30 MPa	Max Stress	: 33.29 MPa	Max Stress	: 40.10 MPa
	Max Deformation Min	: 0.01 mm	Max Deformation	: 7.54 mm
	Factor of Safety	: 7.20	Min Factor of Safety	: 6.02
Acrylic Medium- High Impact	Max Stress	: 32.60 MPa	Max Stress	: 39.90 MPa
	Max Deformation Min	: 0.66 m	Max Deformation	: 0.51 mm
	Factor of Safety	: 1.40	Min Factor of Safety	: 1.14
PE High Density	Max Stress	: 31.79 MPa	Max Stress	: 39.90 MPa
	Max Deformation	: 1.88 mm	Max Deformation	: 1.41 mm
PC High Viscosity	Max Stress	: 32.07 MPa	Max Stress	: 39.9 MPa
	Max Deformation	: 0.86 mm	Max Deformation	: 0.65 mm

b) Rack

During locking, the mechanism will trigger both gear grippers at the left and right sides of each segment to grip the teeth of the rack, thus preventing unnecessary rack movement. For the simulation of the working principle and FEA analysis, the force is applied nearly at the end of the rack. The results obtained were shown in Figures 5, 6, and 7.

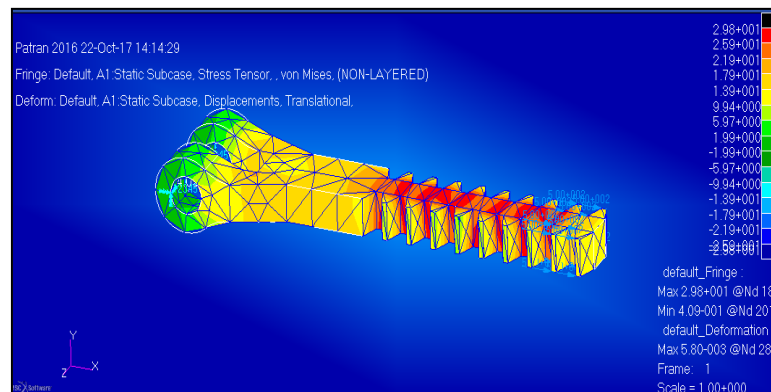


Figure 5 FEA analysis result for stress from Patran-Nastran using TET10

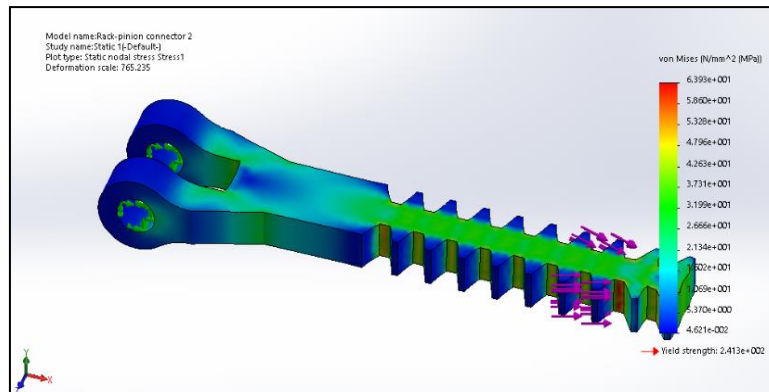


Figure 6 Stress result from Solidworks for Rack

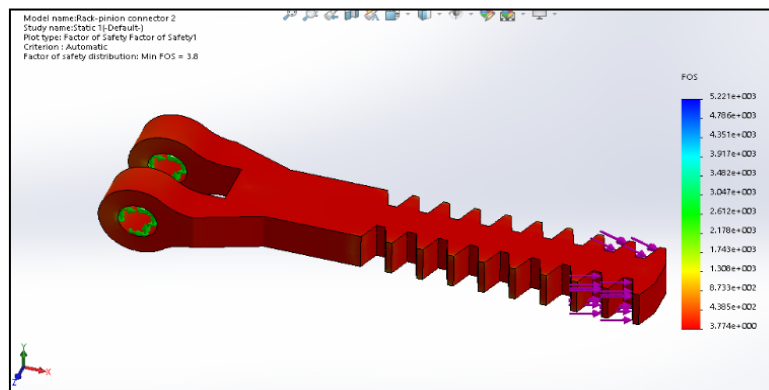


Figure 7 Factor of Safety result from Solidworks for Rack

c) Gear Gripper

From the results obtained, the initial material chosen for the analysis, which is Cast Steel Alloy, failed to sustain 500 N of load during loading conditions as shown in Table 2 below. The peak stress recorded is 343.00 MPa, which is greater compared to the yield strength of the Cast Steel Alloy which is only 241.30 MPa. The geometrical issue may be the reason for the failure as the region to sustain the compressive load is too small in size. Alloy steel, which has better yielding strength, is chosen for the second stage of FEA analysis. And from the analysis results, this material can withstand the applied load as in Figures 8 and 9.

Table 2 Summary of results of FEA for gear gripper

Gear gripper (Material)	FEA software used			
	Solidwork 2017		Patran Nastran 2016 Tet 10 (Nodes 1161, Element 606)	
Cast steel alloy σ_Y : 241.30 MPa	Max Stress	: 216.50 MPa	Max Stress	: 343.00 MPa
	Max Deformation	: 8.02 mm	Max Deformation	: 0.02 mm
	Min Factor of Safety	: 1.10	Min Factor of Safety	: FAIL
Alloy Steel σ_Y : 723.83MPa	Max Stress	: 210.70 MPa	Max Stress	: 333.00 MPa
	Max Deformation	: 7.23 mm	Max Deformation	: 0.02 mm
	Min Factor of Safety	: 2.90	Min Factor of Safety	: 2.17

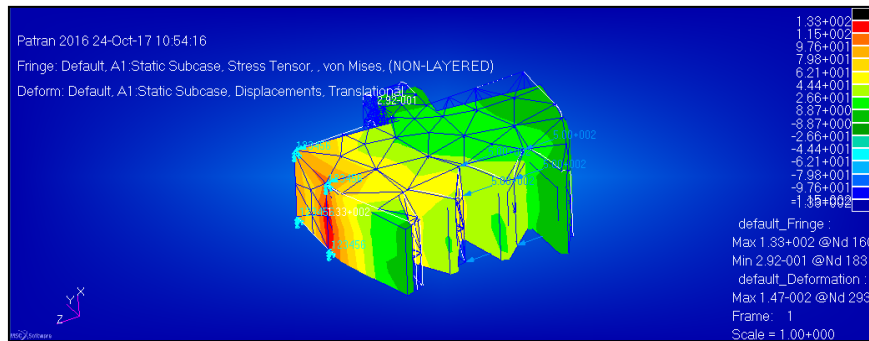


Figure 8 FEA analysis result for stress from Patran-Nastran using TET10 gear gripper.

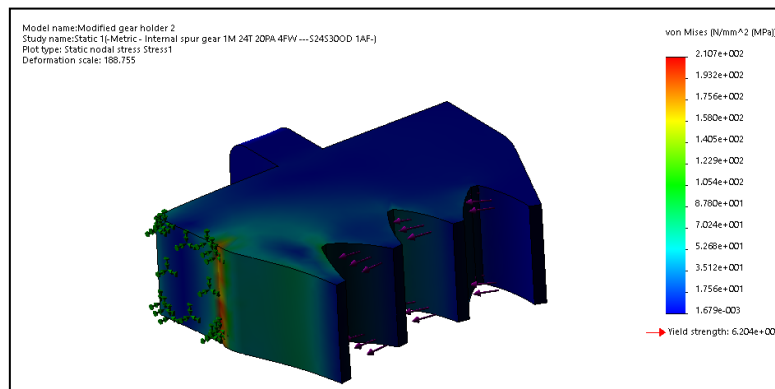


Figure 9 Stress result from Solidworks for gear gripper

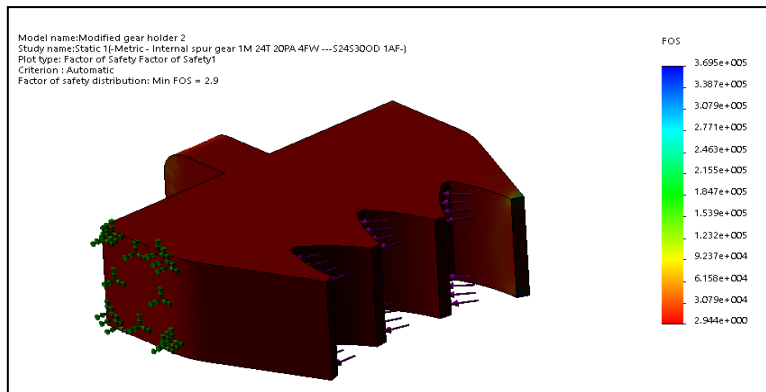


Figure 10 Factor of Safety result from Solidworks for gear gripper

d) Flat-connector

The flat connector allows the rotational motion between both base and rack. Since the dimension of this component is small, stronger material should be considered and used to prevent failure when in use. The results of the FEA analysis were shown in Figures 11, 12, and 13.

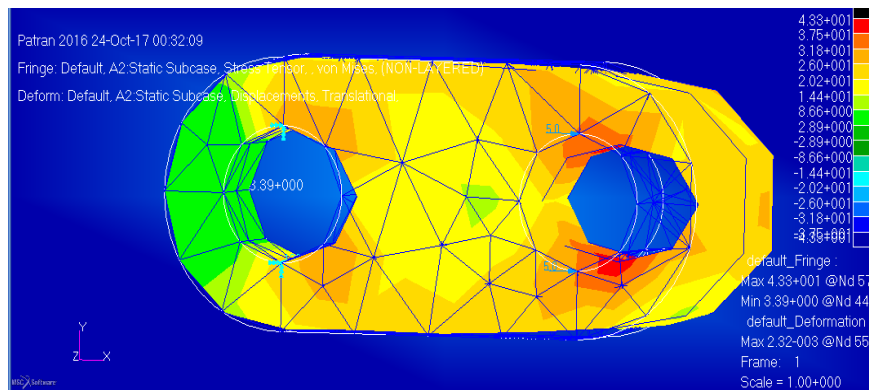


Figure 11 FEA analysis result for stress from Patran-Nastran using TET4 for

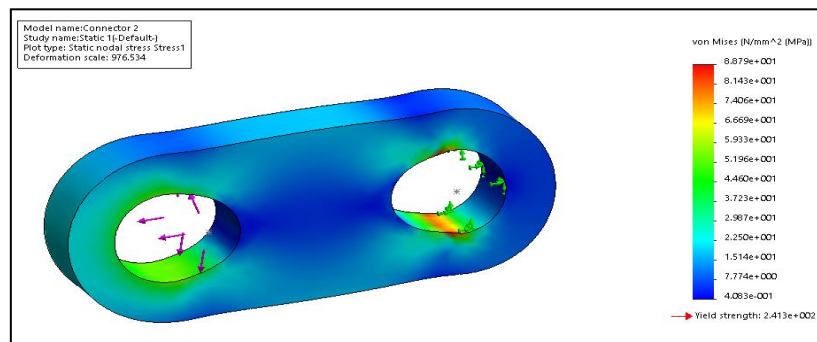


Figure 12 Stress result from Solidworks for flat-connector

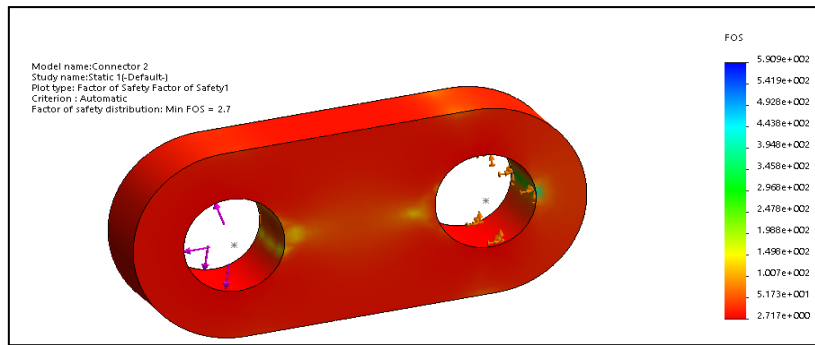


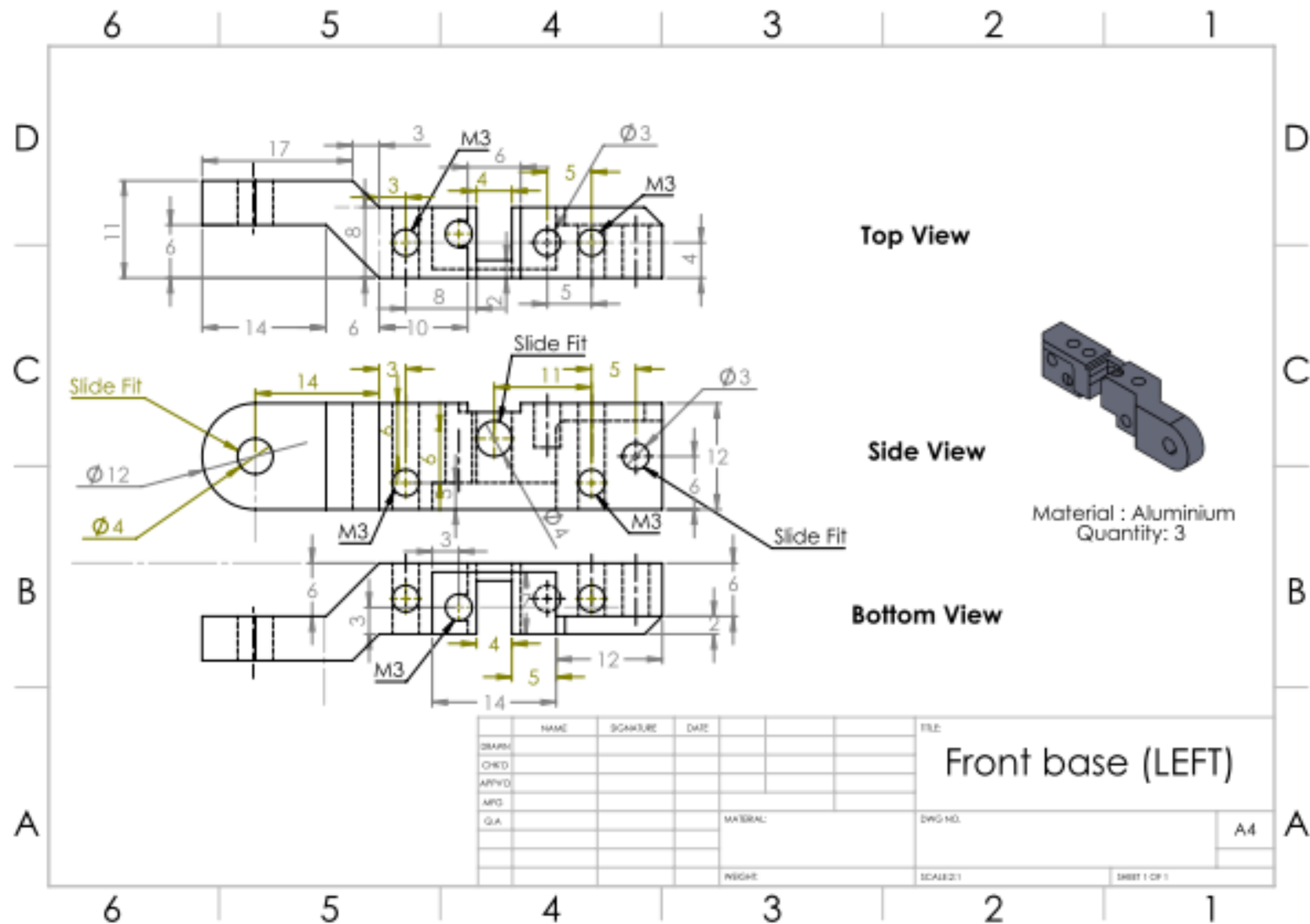
Figure 13 Factor of Safety result from Solidworks for flat-connector

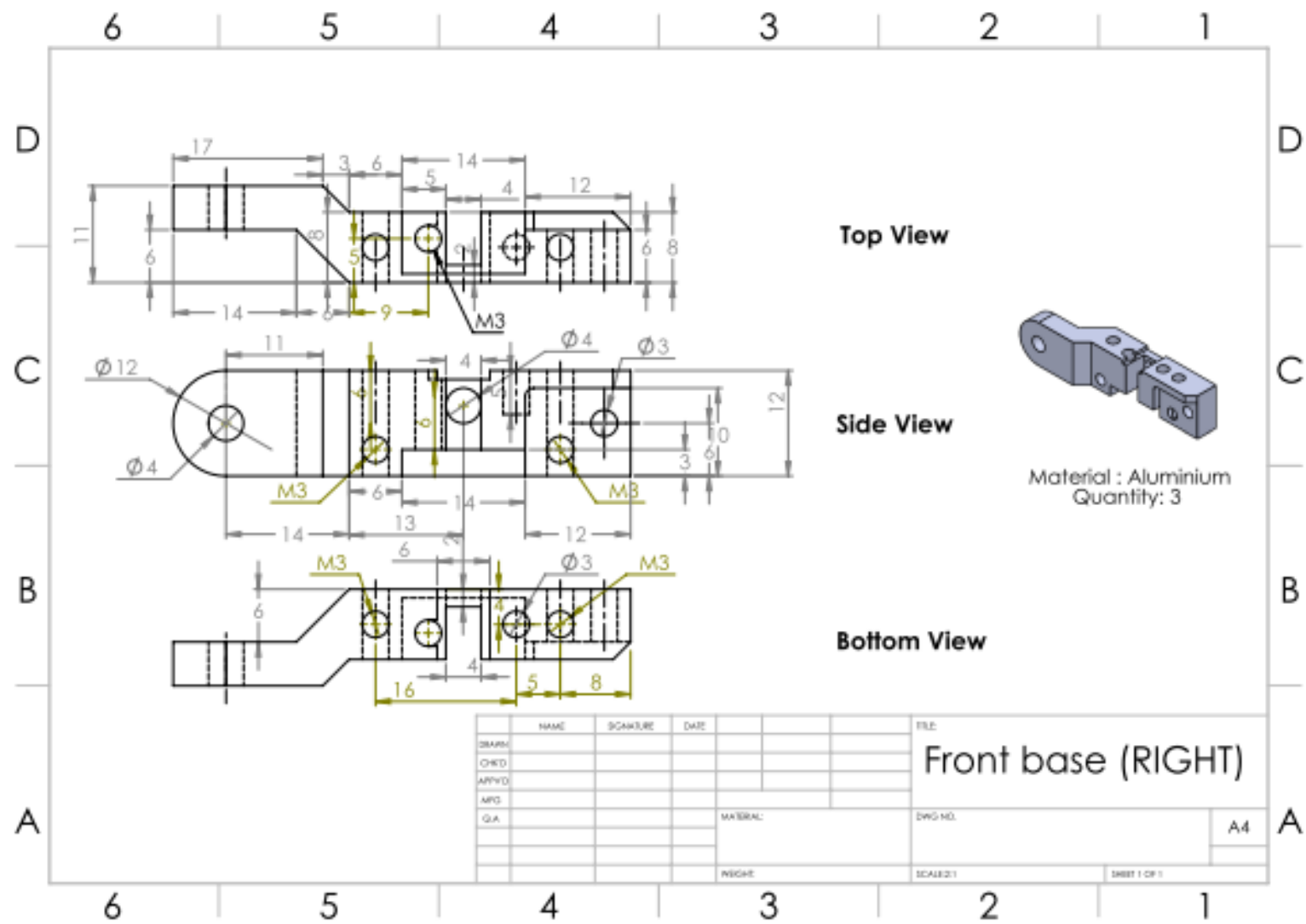
From the results, Cast Steel Alloy failed to sustain 500 N load during operation as shown in Table 3. The highest stress recorded is only 119.20 MPa compared to the yield strength of the Cast Steel Alloy, which is 88.8 MPa. The new material assigned is Alloy Steel which has better yield strength.

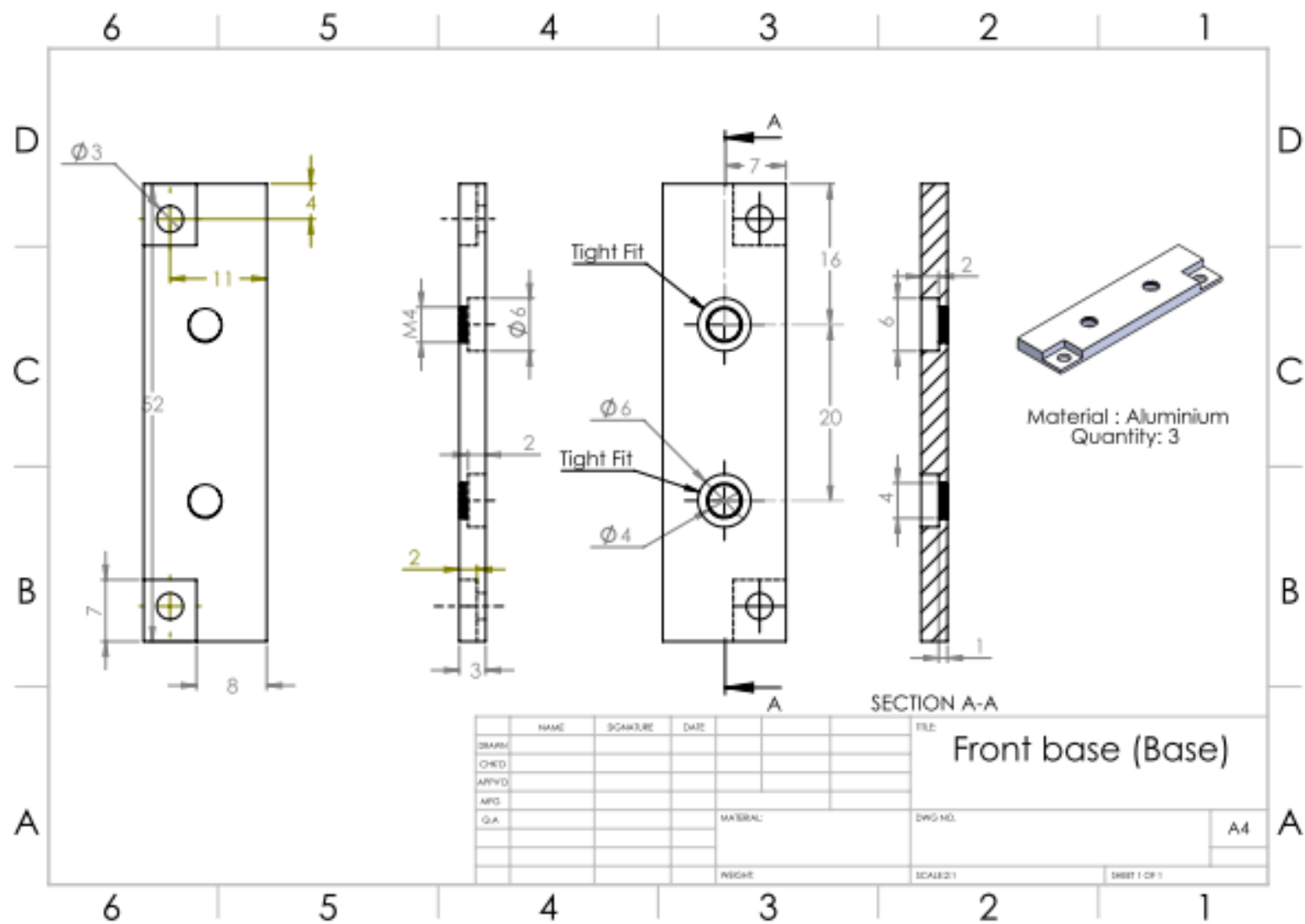
Table 3 Summary of results of FEA for flat-connector

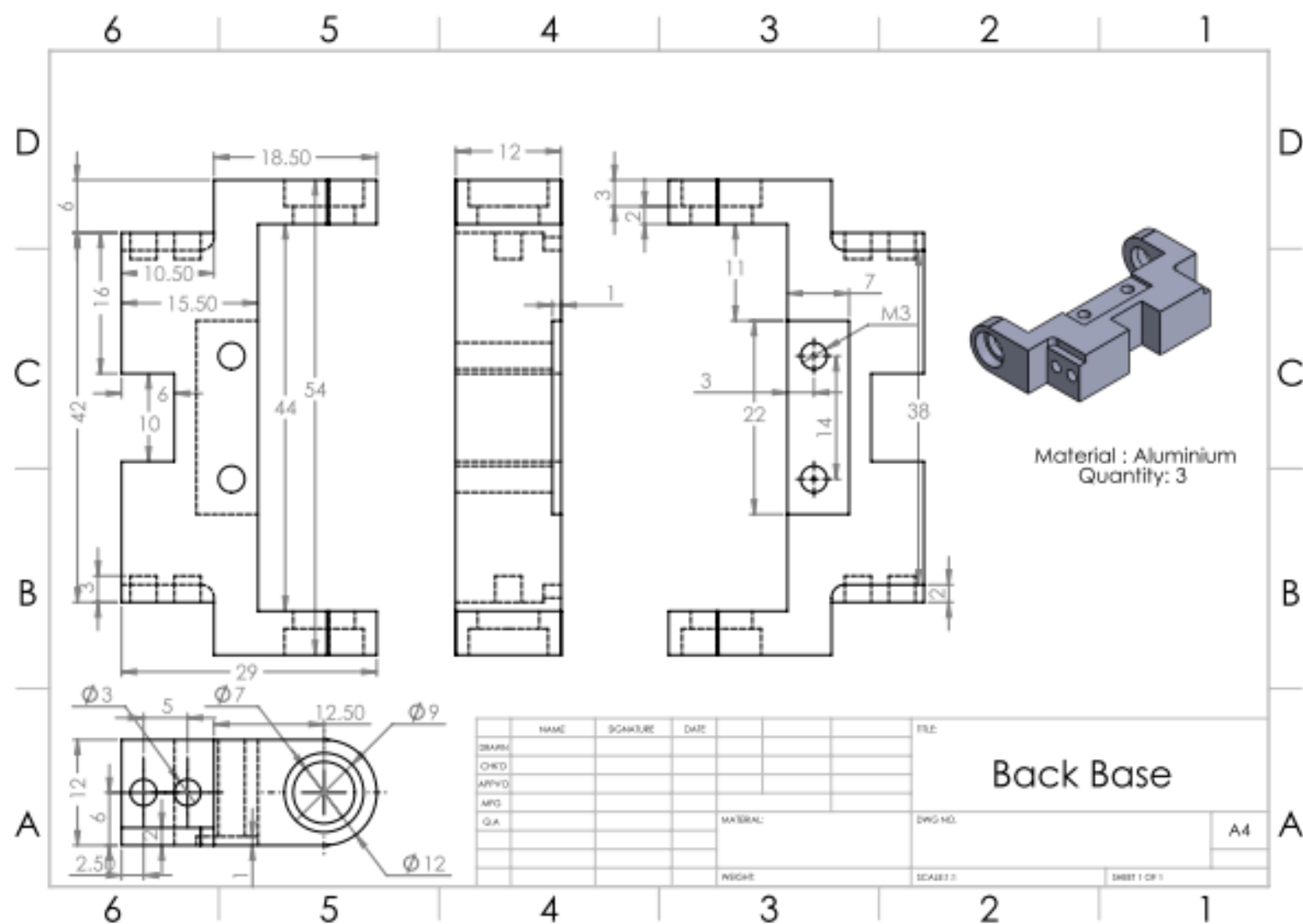
		FEA software used	
Flat-connector (Material)		Solidwork 2017	Patran Nastran 2016
Cast steel alloy σ_Y : 241.30 MPa			Tet 4 (Nodes 156, Element 444)
			Max Stress : 43.30 MPa
			Max Deformation : 2.32mm
			Min Factor of Safety : 5.57
	Max Stress		
	Max	: 88.80 MPa	
	Deformation	: 7.19 mm	
	Min Factor of Safety	: 3.80	
			Tet 10 (Nodes 905, Element 436)
			Max Stress : 119.2 MPa
			Max Deformation : 3.27 mm
			Min Factor of Safety : 2.02

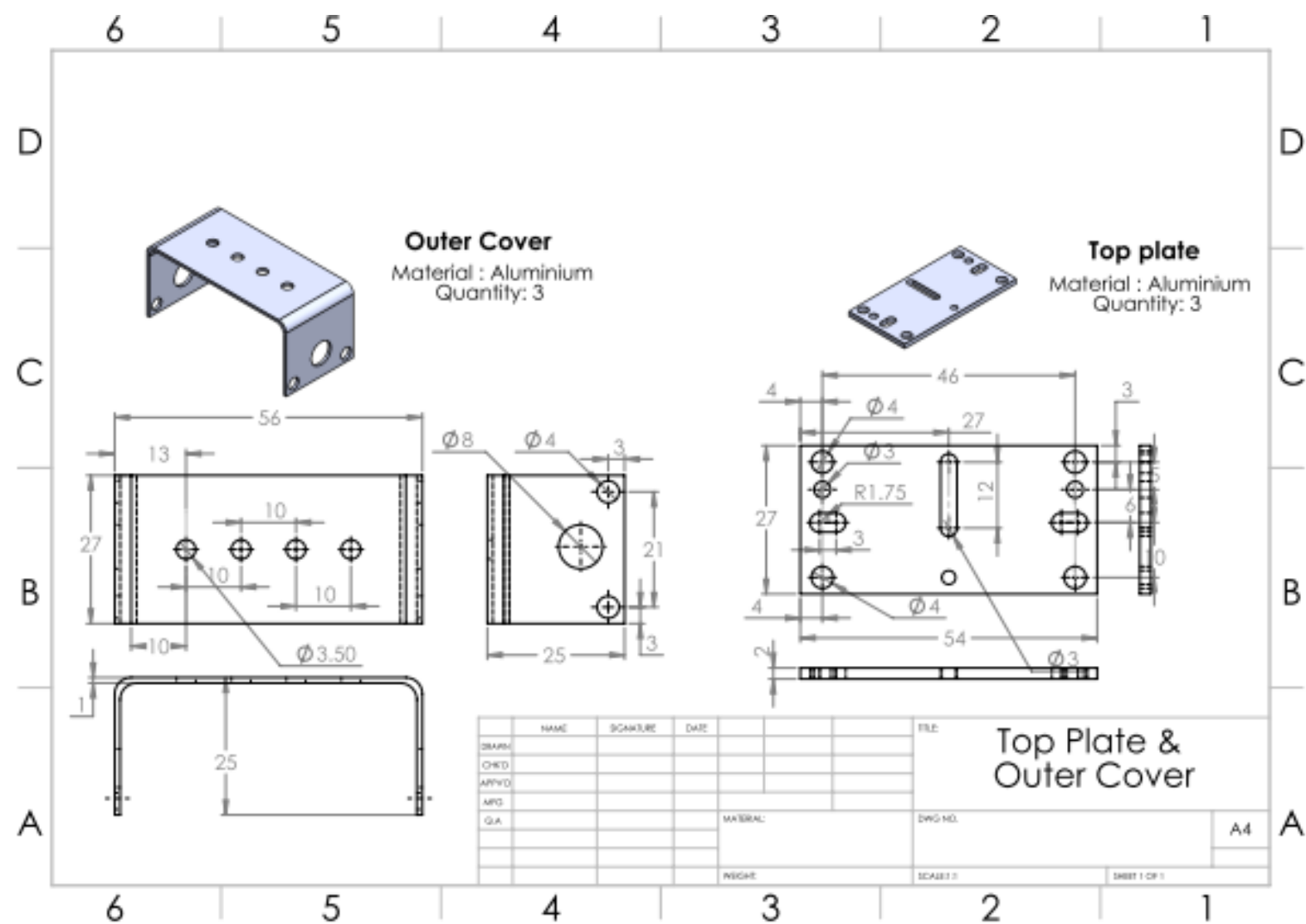
Appendix N Schematic Drawing for Final Design

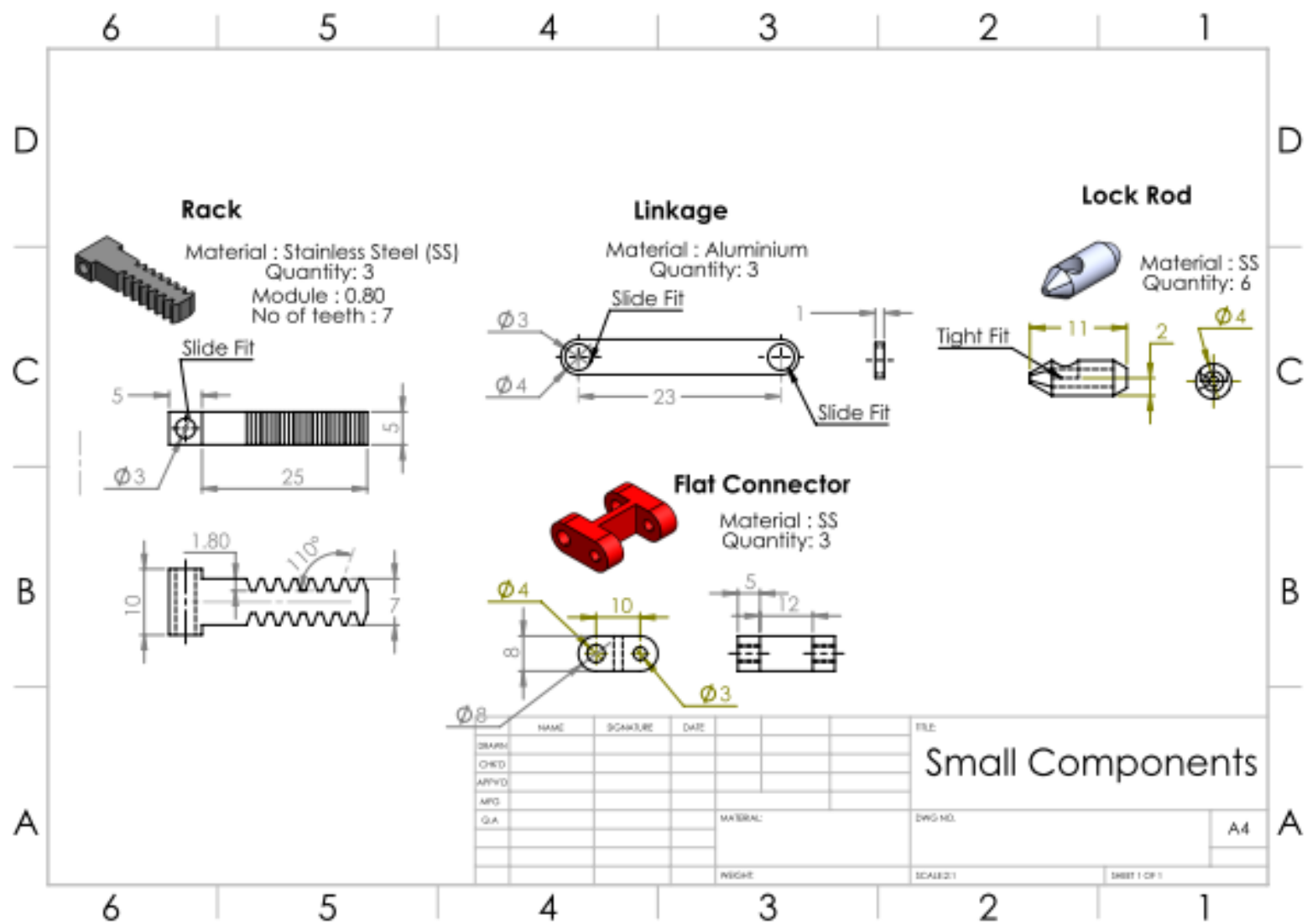












LIST OF PUBLICATIONS

Indexed Journal

1. Saad, W. A. A., Dzahir, M. A. M., Hussein, M., Ahmad, Z. A., Mohamad, M., Saad, S. M., & Dzahir, M. A. M. (2017). Posture Detection on Active and Inactive Vertebrae of Spine Biomechanics in Critical Daily Activities. **(Indexed by Scopus)**

Indexed Conference

2. Saad, W., Dzahir, M. M., Yamamoto, S., Hussein, M., Mohamad, M., Saad, S., & Azaman, A. (2020). Differences of spinal kinematics contribution between cervical and multi-segmental thoracic spine during Sit-To-Stand (STS) & Stand-To-Flexion (STF). Paper presented at the IOP Conference Series: Materials Science and Engineering. **(Indexed by Scopus)**

Conference Proceeding

3. Saad, W. A. A., Dzahir, M. A. M., Hussein, M., Mohamad, M., Saad, S. M., & Dzahir, M. A. M. (2018). Implementation of Stick Marker-Based System in Analysing Kinematics of Kyphosis and Lordosis During Sit to Stand in Healthy Adults.