UNIVERSIDAD POLITÉCNICA DE MADRID

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**Analysis of the Physical Behavior of Spine Joints:
Application / Development of Complex Structures**

TESIS DOCTORAL

Neda Salsabili

Máster en Architectural Technology de University of Tehran

2019

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DOCTORAL THESIS

Neda Salsabili

**Master in Architectural Technology from the University of
Tehran**

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DEDICATION

Dedicated to Merciful Lord who created this world, my beloved parents who were the means of my being in this world, my husband who is my sweetheart in this world, my son who is the fruit of our lives in this world, and my professors who helped me achieve my goal and bring this work into this world.

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RESUMEN

La arquitectura biónica, como parte principal de la Biónica, trata de diseñar y construir edificios basados en diseños y líneas de formas naturales imitando los principios geométricos de la naturaleza. En el siglo XXI, el movimiento de la arquitectura biónica maduró al incorporar los principios biológicos, matemáticos y mecánicos de la naturaleza en el diseño arquitectónico y evolucionó para ser conocido como arquitectura derivada de la naturaleza. En particular, la importancia de este campo se convierte en primordial en una era de crisis ecológica y una necesidad de adaptabilidad y flexibilidad. En este sentido, la forma dentro del curso de la evolución natural, las formas flexibles y la estructura que han sido capaces de rectificar a través de los cambios de la evolución natural demuestran ser las fuentes de inspiración más poderosas y eficientes.

Existen diferentes tipos de criaturas naturales para modelar en el campo de la arquitectura biónica. Esta tesis se centró en la articulación de la columna vertebral como un buen ejemplo de estructuras naturales flexibles y adaptadas que son similares a otras articulaciones humanas, al tener estas seis características clave, 1- forma estética al armonizar en respuesta a las fuerzas ambientales, 2- complejidad, facilitando simultáneamente para funciones de flexión, extensión, abducción, aducción, rotación y circunducción, 3- comodidad al asegurar la estabilidad en diferentes posiciones, 4- dinámica al facilitar el equilibrio a través de movimientos flexibles y fuerzas internas y externas, 5- sistema estructural de luces y 6- forma modular por su estructura repetida. Los roles desempeñados por la articulación de la columna vertebral son muy similares al tipo de rol requerido dentro del diseño industrial y de la construcción, por lo que es un elemento natural que puede informar el futuro de la arquitectura y el diseño industrial. Esta simulación se realiza mediante el uso de la combinación de programas y software de biología, construcción, matemáticas, mecánica y computación. Sin embargo, la simulación de dicha estructura requeriría una simplificación de la complejidad estructural y material de la articulación de la columna vertebral de acuerdo con los recursos arquitectónicos y de construcción disponibles. Esto, a su vez, también facilitaría el proceso de modelado de tales estructuras para la bioingeniería y la construcción de prótesis, así como otras estructuras.

El método de elementos finitos que se utiliza en esta investigación al usar un rango diferente de propiedades es bueno para modelar diferentes partes geométricas y puede ayudar a simular, analizar e investigar el efecto de simplificación de las partes materiales de la columna lumbar humana de una manera más fácil y más fiable, evitando el daño a la columna vertebral, reduciendo el tiempo, las pruebas, los prototipos físicos y los usos del material, mejorando la seguridad y la información, optimizando la geometría diferente para las prótesis y las diferentes condiciones fisiológicas, midiendo diferentes distribuciones de tensiones, desplazamientos y mecanismos de transferencia de carga, y finalmente, ayudando a desarrollar los nuevos implantes espinales. Por lo tanto, ofrece ventajas en comparación con los métodos in vivo que tienen limitaciones en términos de medición, obtención de la muestra y la variación de la muestra, son costosos y se ha informado que consumen mucho tiempo.

Para hacer las nuevas estructuras basadas en la columna lumbar humana al tener sus propiedades anatómicas y mecánicas en el método de FE para la construcción de uniones y

estructuras adicionales, en esta tesis se utilizó una metodología basada en cuatro fases: primero se hizo la FE de madera utilizando una combinación de software diferente con los datos de tomografía computarizada, fórmula verificada y técnicas para calcular el modelo, y la combinación de materiales, cargas y propiedades geométricas de los datos del modelo validado, en segundo lugar, se eliminaron sus materiales en cuatro pasos continuos, en tercer lugar, despojado de sus estructuras en cuatro pasos continuos también, y finalmente fue escalado 10 y 100 veces más para comprobar su sensibilidad para modelar las juntas del edificio y la estructura del edificio. El modelo en cada parte de la metodología se validó al comparar sus datos de presión intradiscal (IDP) y rango de movimiento (ROM) con datos experimentales in vitro e in vivo.

Hemos demostrado que los resultados de estas estructuras de elementos finitos en desarrollo fueron consistentes con los datos experimentales de IDP y ROM in vivo e in vitro en todas las direcciones y en todos los niveles de desarrollo, simplificaciones y escalado. Además, la nueva junta y estructura del edificio puede tolerar cargas de compresión de 10^2 y 10^4 veces, y los momentos de 10^3 y 10^6 veces más grandes que el tamaño normal en 10 y 100 veces la escala respectivamente en cada dirección con una buena reducción en la cantidad de estrés.

El método de modelado validado introducido en este estudio se puede utilizar para futuras investigaciones de fabricación de implantes y prótesis, modelado basado en la columna lumbar humana para futuras juntas de construcción (10 veces a escala), diseño industrial, estructura de construcción (100 veces a escala) y alto - edificios de gran altura debido a la forma modular de la columna vertebral humana en su sección lumbar con su capacidad de tolerar estas grandes cantidades de cargas en diferentes direcciones, teniendo la respuesta interna y externa a diferentes tipos de cargas y siendo adaptables a la naturaleza.

ABSTRACT

Bionic architecture as a main part of Bionics tries to design and construct the buildings based on layout and lines of natural forms by mimicking the geometrical principles in nature. In the 21st century, bionic architecture movement matured by incorporating the biological, mathematical and mechanical principles of nature in architectural design and evolved to be known as architecture derived from nature. Notably, the significance of this field becomes paramount in an era of ecological crisis and a need for adaptability and flexibility. Respectively, form within the course of natural evolution, flexible forms and structure that have been able to change through the changes of natural evolution prove to be the most powerful and efficient sources of inspiration.

There are different types of natural creatures for modeling in the field of bionic architecture. This thesis focused on the spine joint as a good example of flexible and adapted natural structures similar to other human joints, since they have these six key features, 1- aesthetic form by harmonizing in response to environmental forces, 2- complexity, by simultaneously adapting to functions of flexion, extension, abduction, adduction, rotation, and circumduction, 3- comfort form by remaining stable in different positions, 4- dynamic form by facilitating balance through flexible movements and internal and external forces, 5- light structural system, and 6- modular form by its repeated structure. The roles performed by the spine joint are very similar to the type of role required within the building and industrial design and therefore it is a natural element that can inform the future of architecture and industrial design. This simulation happens by using the combination of biology, construction, mathematics, mechanics, and computational programs and software. The simulation of such structure, however, would require a simplification of both in the structural and material complexity of the spine joint in accordance with the architectural and construction resources available. This, in turn, would also ease the process of modeling of such structures both for the bioengineering and construction of prosthetics as well as other structures.

The Finite element method (FEM) which is used in this research by using a different range of properties is good for modeling different geometrical parts and it can help to simulate, analyze, and investigate the effect of simplification of the material parts of human lumbar spine easier, without damaging on spine, reducing the time, the testing, the physical prototypes, and the material usages, improving the safety and information, optimizing different geometry for prosthesis and different physiological condition, measuring different distribution of stresses, displacements, and load transferring mechanisms, and assisting to develop the new spinal implants. Therefore it offers advantages in comparison to the vivo methods which have limitations in terms of measuring, obtaining the specimen, the specimen variation, being costly, and have been reported to be time-consuming.

In this thesis, to make the new structures based on human lumbar spine by having its anatomical and mechanical properties in FE method for further building joints and structures, a four-step methodology was used: first the FE of lumbar was made by using a combination of different software with the CT scan data, verified formula and techniques for calculating the model, and the combination of materials, loads, and geometry properties from the validated model's data,

secondly it was stripped of its materials in four continuous steps, thirdly it was stripped of its structures in four continuous steps too, and finally it was scaled 10 And 100 times greater to check its sensitivity to model the building joints and the building structure. The model in each part of the methodology was validated by comparing its intradiscal pressure (IDP) and range of motion (ROM) data with vitro and vivo experimental data.

We have shown that the results from developing structures from the FEM were consistent with the experimental IDP and ROM vivo and vitro data in all directions and in all level of developing, simplifications, and scaling. Moreover, the new building joint and structure can tolerate compression loads of 10^2 and 10^4 times, and the moments of 10^3 and 10^6 times larger than normal size by 10 times and 100 times scaling respectively in each direction with a significant reduction in the amount of stress.

The validated modeling method introduced in this study can be used for future research, in particular in producing implants and prosthesis, modeling based on the human lumbar spine for future building joints (10 times scaled), industrial design, building structure (100 times scaled), and high-rise buildings due to the modular form of the human lumbar spine with their capacities of tolerating against these large amount of loads in different directions, having the internal and external response to different types of loads, and being adaptable to nature.

CHAPTER I: INTRODUCTION

1.1 Background

Bionic aims to model from nature in order to solve human problems, it doesn't seek to copy or imitate nature. The process of creating natural forms has had a long evolution over millions of years unconsciously and during this process, the most efficient, powerful and flexible natural forms have been able to survive (Salsabili, Mahmoudi, Salsabili, & Nasr Azadani, 2016). So the solution for ecological crisis is provided by nature. Therefore, the urgent need to achieve environmental sustainability in the early 21st century causes to promote (Versos & Coelho, 2011) this science as a conscious modeling (Salsabili, 2014) and become as one of the three world's central sciences (Bionics, Nano, IT) (Salsabili et al., 2016).

Bionic architecture as a main part of Bionics tries to design and construct the buildings based on layout and lines of natural forms. Inspiration from nature in architecture has existed from time immemorial. In the beginning, it was in the development of aesthetic and economic shape and then, it is faced the combination of realities based on biology, mathematics, construction, and mechanics (Salsabili et al., 2016). Bionic architecture in a general classification can be divided into 6 main categories of stencils, structure, materials, form, function, and process (Salsabili et al., 2016).

Nature is an unlimited source of inspiration for bionic architecture and a good teacher in various fields of knowledge, technology, science and architectural design (Taghizadeh & Bastanfard, 2012). Modeling has forever been present in architecture but many times unconsciously. The early 21st century witnessed the birth and maturity of a movement combining biology, mathematics, construction, and mechanics. This approach is called architecture derived from nature (Salsabili, Barrio Prieto, & Santiago López, 2018). Each natural element is the complex of several systems, and each of these natural systems consists of a number of subsystems (components and materials) which are working as separate parts too. This relationship between human and nature divided into four stages:

- Nature and organic model (Hunting Age or Stone Age), a human being has been overcoming nature in all positions of life.
- Nature and hyper-organic model (the farm age), trying to be comforted by nature of the sense of "being with nature".
- Nature and dominant nature model (the industrial revolution), having the maximum utilization of nature by a human.
- Nature and industrial quasi-organic model. Nowadays, the global community is trying to adapt industry to nature (Taghizadeh & Bastanfard, 2012).

The human body is a good example of nature for the near-future architecture because of its shapes, aesthetic forms, features, characters of forms, light form, comfort forms, and efficiency and dynamic behavior that actively have to respond its forces by having its stability and balance while moving. Based on biomechanics, the human body is a complex and variable set of levers to balance the body against the external forces (Taghizadeh & Bastanfard, 2012). It contains (Salsabili et al., 2018):

- Joints are formed at the junction of two or more bones to make the framework of the body. Different types of joints provide various degrees of movement (Taghizadeh & Bastanfard, 2012) such as flexion (decreasing joint angle), extension (increasing joint angle), abduction (movement away from body center line), adduction (movement toward body center line), rotation (a circular movement around a fixed point), and circumduction (360 degree rotation).
- The brain and nervous system coordinates actions between different cells and organs of the body, control and send a message to muscles.
- Muscles (640 numbers in human bodies (40% of human body weight)) have different shapes and size to provide stability and ability to do many functions such as moving, breathing, running, walking, writing and dancing (Salsabili et al., 2018; Taghizadeh & Bastanfard, 2012).

Human Joints as a starting point are highly beneficial for modeling in architecture on account of having different types of activities such as flexion, extension, abduction, adduction, rotation, and circumduction in different positions such as bending, lying, bridging, running and moving with its muscles and brain system. On the other hand, it is stable in the face of different outdoor forces without muscles such as hitting, impacting and bearing the load. So according to these types of joints, we could make different types of structures that have the best ability to be resistant according to outdoor forces such as wind or earthquake or to be resistant in different positions such as bridge, high rise buildings and rotary structure (Salsabili et al., 2018).

Skeletal system, as a source of modeling, is categorized into three types of immovable joints, slight movement joints and full movement joints based on their functions and motions. Besides, it consists of three groups of synovial joints, fibrous joints, cartilaginous joints based on their structures. These types of joints categorized into subgroups and each subset has a specific character for simulating. Different types of these structures were designed consciously or unconsciously, more or less similar to human joints up to now. But knowing the function and structure of these types of joints can help architects to design buildings more effective and sustainable (Salsabili et al., 2018).

1.2 Problem statement

From the beginning of the human existence, the relation with nature solves the challenges of building with details such as mechanical, electrical, structural, protection, cooling, shell building, and security by making natural shelters or archetypes by using caves and trees. However, humans used nature and the environment unfeeling without paying attention to the negative effect their actions could have on the planet and thus upsetting the planet's balance by causing global warming and climate changing. Nowadays, Bionics structures are modeled based on reviewing and studying the structures and lines of nature, which has evolved over millions of years based on human needs. Therefore, these structures bridge the gap between nature and technology and cause a revolution in the development of technology. It is still being developed (Taghizadeh & Bastanfard, 2012) by the combination of realities based on

biology, mathematics, construction and mechanics to make algorithmic geometry, fractals theory, chaos theory, intelligent materials and systems, opening and closing mechanisms (Salsabili et al., 2016), genetic algorithms, algebra Fractal, and neural networks. This development causes sustainable development of complex systems by increasing the efficiency of resources (Al-Obaidi, Ismail, Hussein, & Rahman, 2017), thus; the more similar our performance is to nature, the longer human is survival. On the other hand, humans are using nature and the environment recklessly and thus have upset the planet's balance.

There are extensive examples in nature for modeling and many of them have already been used, but taking the advantages of the human body especially in the field of construction has been neglected. At first glance, it might be thought that the human body is very complex, but by studying this point, we can conclude that in spite of our imagination, it is much understandable and usable structure, there have been many studies about its' behavior also.

Nowadays, due to the mentioned reasons, it becomes important to study the principles of human joints. The human joints (the frame of the human body) are the good examples of modeling because of their light structures with both opposite aspects of stability and mobility that they have. On the other hand, they are more different with respect to each other in their roles of moving as their functions. Therefore, they are good examples of modeling to make different structures with different capacity and functions to be resistance against the wind and earthquake.

1.3 Preliminary hypotheses

According to what has been presented in the previous sections, we aim to focus our analysis on the human joints, because of the new freedom of creating by the recent advances in materials, construction techniques, calculation methods and structures based on these joints factors (Taghizadeh & Bastanfard, 2012) such as:

- Joint shapes and aesthetic form: shapes in nature are formed for the best function and using the least material which is the result of compromising with the environmental powers to make synthetic forms and systems ("Types of Joints," n.d.; Zakharchuk, 2012).
- Joint features and characters: the complex roles of different functions (flexion, extension, abduction, adduction, rotation, and circumduction) that each joint have in stability and mobility make it a good sample of modeling more than its form (Zakharchuk, 2012).
- Joint as a stable and comfort form: the human body can be stable in different parts of its joints in a different position of lying down, standing, leaning, bridging, etc. while moving.
- Joint with the dynamic form: one of the features of the joint is their ability to move and change to be balanced in its different positions of running, walking, jumping, crawling, and etc. while moving (Zakharchuk, 2012).
- Lightweight structure: the percentage of its weight per scale is very low.

- Modular form: the spine contains numbers of vertebrae, which are attached to each other by an intervertebral disc, ligaments, and facet joints and can be repeated by the modular form.

By having several types of joints in our body, this study has focused spine joints as a sample. Throughout this research, it has been a struggle to answer fundamental as well as practical questions regarding modeling, calculating, validating, improving spine structure and materials to make a light and dynamic structure to resistance against the wind and earthquake.

Furthermore, through the kinetic study carried out in this research, scientists would be enabled to have a clearer and brighter overview of the making structure based on human body joints as a benefit structure.

1.4 Relevance of the study

The purpose of this study is to extend the understanding of human spine structures (shape, anatomical and mechanical behavior, structures, and material) in order to make new structures for bridges, high-rise buildings, connectors, and a model for making moveable structures with stable parts.

As stated in previous section, because of its anatomical and mechanical behavior, the human spine can play an important role in making modern structures. Moreover, the outcome of this thesis can be used by other researchers in the fields of industrial design, making prosthesis or implanting, etc.

Everything considered, this work aims to achieve the following objectives, which are divided into two parts of major objectives and minor objectives:

Major objectives

- Designing the building structure based on the anatomical behavior of the human lumbar spine, which is tolerating against the internal and external loads.
- Making Finite element (FE) of human spine based on mechanical and anatomical behavior of the human spine through FEM based on its CT scan and verified data by compound of software.
- Validating the FE of human spine by comparing with the experimental vivo and vitro data. Moreover, checking the sensitivity of the FE of the human lumbar spine by scaling it to the 10 and 100 times greater.
- Designing the simple structure by simplification the material part of the human lumbar spine to make further structures for implanting or prosthesis or knowing the mechanical behavior of each material to the internal and external response.
- Creating the elemental and basic structure based on human lumbar spine by simplification both materials and structural parts for further joints in industrial design or building structures.

- Testing the sensitivity and characters of the model by scaling the new structure based on the human lumbar spine 10 times or 100 times greater to understand the new structure capacity to make the building joints and building structure.

Minor objectives: besides the major objectives, there are some minor objectives, which have to be done and studied.

- Studying about bionics, bionic advantages, bionic approaches, bionic steps for modeling, and the most examples.
- Defining bionic architecture as a main part of bionic science with its types, pioneers, structures, history, level of mimicking, and its interaction with nature by setting the main examples of that.
- Defining nature and the human body relation with it.
- Studying about the human body by its structure and finding the relation between human body, architecture, building design, and future structure. Moreover, specifying why the human body is a good solution for future building.
- Understanding the skeletal systems by knowing its main parts such as joints, muscles, and spinal cord and nerves.
- Knowing the types of joints based on movement, structure, the degree of freedom, shape, biomechanical classification, anatomical classification, and their actions, which are useful for further modeling.
- Determining why different types of skeletal joints may be useful for making the future joints and structures in a different field of industrial design, building structures, medical prosthesis or implanting by describing their anatomical roles.
- Describing the structures, joint, or building, which were made based on human joints consciously or unconsciously, and categorizing them according to the human joint types.
- Studying the anatomical behavior of human spine, which is, contained the vertebrae, intervertebral disc, ligaments, and facet joints.
- Studying the mechanical behavior and properties of human lumbar spine such as its materials, characters, scale and dimension of each part of the human lumbar spine through the mechanical resources.
- Setting the main chart of human lumbar spine mechanical behavior, which is adaptable and useable for further steps of modeling by software or making prototype.
- Describing the type and amount of loads for each direction of loading for the human lumbar spine.
- Analyzing the validated experimental vivo and vitro examples data for setting the ranges for validating, analyzing, and comparing our future data.

- Describing the Finite Element method (FEM), its benefits, its problems, its steps of modeling, and its related software.

1.5 Significance of the study

Making building structures and building joints based on human joints and spine is new and the contribution of the current study would be of great interest to scholars in architecture, engineering, industrial design, building, and biomechanics by making a new, light, and sustainable structure in different positions because of the potential of human spine structure.

The core areas of the present research are in the field of innovation in the building to which my study would be significant. Throughout this research, the anatomical and mechanical behavior of the human spine would be investigated and discussed in details to find the structures based on that. Then promoting these structures by software to make new structures, which are resistance against a large number of compressions and moments.

This study must be used for researchers who want to make new and sustainable structures based on nature. This study shows the line of modeling from the natural form to make the building structures and real form.

1.6 Thesis structure

This thesis contains seven parts of introduction, literature review, materials and methodology, results and discussions, conclusions and recommendations, bibliography, and list of tables.

Some parts consist of more than one chapter, so the whole thesis consists of eleven chapters with bibliography and tables. Each chapter is developed gradually to receive the whole data, which are proper to develop the thesis.

Introduction

This part contains only one chapter of Introduction, which is described below:

Chapter I: Introduction

It contains subsets as follow:

- Background information (Review of the literature).
- Problem statement (The gap in this research, justification, and defense of the choice).
- Research questions (Main objectives and aims of this work, need for this study and contribution to architectural design).
- Objectives of the study (Improving steps of the working hypothesis).
- The significance of the study.

- Thesis plan (Brief description of the structure of this work, its chapters and sections, and the methodological approach adopted, plus materials means and resources used, the way of developing the structures, as the case may be).

Some of the terms used in the thesis plan have been briefly defined which will be discussed in detail in the subsequent chapters.

Literature review

This part contains four chapters. The first one is about bionic and its approach in the field of architecture. The second one is about the skeletal system of the human body as a good source of nature for modeling. The third one is about the anatomical behavior of human lumbar spine as a part of the human spine and presenting a model based on that. The last one is about the mechanical behavior of the human lumbar spine.

Chapter II: Bionics

The aims of this chapter are to describe the general background of bionic and its literature in the field of architecture and nature as a good example of modeling and the human body as a good example of nature, which is neglected in architecture.

Chapter III: The skeletal system

The aim and purpose of this chapter are to describe a skeletal system of the human body and its joints as a framework of the human body to set the theoretical framework for the study. The detailed explanation of human body joints and their structures that we are faced today which are consciously or unconsciously designed and worked.

Chapter IV: The anatomical roles of the human spine

This chapter consists of a detailed description of the anatomical behavior of human spine, which leads to making a model based on them.

Chapter V: The mechanical roles of the human spine

The aim of this chapter is to know about the mechanical behavior of each part of the human spine and its relations with the rest of the skeleton at the level of stress, deformations, properties of materials, types of knots, behavior ... with its charts of mechanical behavior, which are made the vertebral joint. This chapter helps to make better modeling of the human spine and FE by software.

Materials and methods

This part contains three chapters promoting the human spine to receive building structures. A detailed description is presented in these chapters on the Finite Element Method (FEM) to receive the main goal of making building structures.

Chapter VI: The finite element approach

There is a description of the Finite Element methods and its advantages and disadvantages in this chapter. Then the way and approach of making any model by FEM especially the human spine is described.

Chapter VII: Modeling the human lumbar spine by FEM

This chapter describes the levels of making FE of human lumbar spine based on CT scan data of human spine, verified data, which are gained from the mechanical behavior part by using combinations of software. Then the FE of the human lumbar spine was validated by comparing its intradiscal pressure (IDP) and range of motion (ROM) data with experimental vivo and vitro data, and checking its sensitivity in a larger size to understand is it feasible to make structure based on that or not?

Chapter VIII: Simplifying the FE of the human lumbar spine

This chapter describes the way of making the building structures and joints by simplifying of the human spine in material and structural parts by different software and their verified data. Then the new structure was validated by comparing with the normal one data. At the end, it was scaled greater to check this new structure for a bigger size, building structures, and building joints.

Results and discussions

This part consists of two chapters which is contained the data which are gathered from the previous parts and then there is a large discussion about those data.

Chapter IX: Results

This chapter systematically shows all the results gained from all experiments carried out during the study. All results are presented in the form of pictures, charts, and diagrams and the key features are subsequently discussed. This chapter consists of solo pure results gained from solutions sets of the modeling part.

Chapter X: Discussions

This chapter systematically discusses all the results gained from all experiments carried out during the study. All analytical results are presented in the form of diagrams and the key features are discussed. Further discussions are made over the gained results and it has been tried by the author to demonstrate the results in the form of diagrams to make them more understandable. This chapter consists of analyzed results gained from characterization sets along with the results of the modeling part. Validation of results obtained from the experimental phase.

Conclusions and future research lines

This part contains two chapters which are described the main result and general conclusions of this study; moreover, it contains the future recommendation for further study.

Chapter XI: Conclusions

The general conclusions of the study based on the results of carried out experiments and tests plus the model results are presented in this chapter.

Chapter XII: Future research lines

Further recommendations for future studies in related fields are presented in this final chapter.

Bibliography

List of tables and figures

CHAPTER II: BIONICS

2.1 Bionics

Since the beginning of civilization, bionic has been a science that can bridge the gap between technology and nature because it is the combination of nature and the concepts of engineering science (Taghizadeh & Bastanfard, 2012). The word "bionics" is the blend resulting from the two words, biology and technics, biology and technic, which means techno biology. Jack E. Steele at the conference "living patterns" in 1960 applied this word for the first time. So, bionic science deals with the review and study of the structures and patterns in nature and apply them to solve human problems (Salsabili et al., 2016). This science is not seeking to copy or merely imitate nature, but it aims to model the knowledge derived from nature correctly (Salsabili et al., 2016). Although it can be said that all human engineering is a kind of modeling from natural models consciously or unconsciously, this science as conscious modeling does not have a long history (Salsabili, 2014). Finding a solution to the ecological crisis is provided by nature. Therefore, urgency in achieving environmental sustainability in the present days causes to promote a renewed interest in gathering inspiration from nature to make novel design concepts. Finding a benchmark of nature is the main purpose of bionics. Because the things that are created have evolved over millions of years (Versos & Coelho, 2011).

2.1.1 Advantages of inspiring from bionics

Over billions of years from the emergence of nature, bionics has been able to solve the problems, which we are facing today. Animals, plants and unicellular organisms are expert engineers that are well adapted to their environment and show us the way of interacting with nature. What has remained for us in the 3.8 billion years is the lesson that its learning will lead to the reduction of any material costs and prevent waste of resources and time. The more similar our performance is to nature, the more guarantee for our longer survival. Therefore, the process of creating natural forms has had a long evolution journey over millions of years. During this process, only the most efficient, the most powerful and the most flexible natural forms have been able to survive these millions of years (Salsabili et al., 2016).

2.1.2 Bionic-based approaches

Bionic-based approaches as a design process are generally divided into two groups:

- **Problem-oriented approach:** Defining a human need or design issue and studying other organisms and ecosystems to find what solutions and strategies they can use for dealing with this issue. There are six steps in inspiring from nature:
 - Design issues.
 - Searching to find biological similarities.
 - Identifying appropriate legislation.
 - Abstraction and separation from the biological model.

- Testing, analysis, and feedback.
- Design solution.
- **Solution focused approach:** Specifying a particular characteristic, behavior or performance in an organism or ecosystem and translating it to design a human who is called as a biologist influencing design. This approach also has six steps:
 - Biological researchers.
 - Biomechanics, morphological yield, and anatomy.
 - Understanding basic regulations.
 - Abstraction and separation from the biological model.
 - Technical tools.
 - The design solution (Salsabili et al., 2016).

2.2 Bionic architecture

Bionic architecture aims to design and construct buildings based on layout and lines of natural forms. The early 21st century was the time of maturing this movement. The first step was the development of aesthetic and economic justifications for their approach to architecture. Bionic in architecture is an emerging field that is currently being defined and explored. The application of observations made in nature to architecture has always been a challenge for architects and designers (Salsabili et al., 2016). Therefore, inspiration from nature in architecture has existed at all times. The first human dwellings were natural shelters, and archetypes such as caves and trees have been used as models for architectural design throughout history. By the quirk of fate, all architects who have been interested in nature have achieved valuable scientific and theoretical successes (Salsabili et al., 2016).

Bionic architecture is proposed as a style with different and variety of samples for many decades. It is also one procedure of using the structure of living and organic systems in architecture. Although today, in the twenty-first century, we see a version more different than the former definitions of bionic architecture which deals with complex interpretations of the combination of realities based on biology, mathematics, construction, and mechanics and it is called architecture derived from nature. New sciences associated with algorithmic geometry, fractals and chaos theory, intelligent materials and systems, opening and closing mechanisms, all make changes in our landscapes. In fact, our view towards the structure of natural organisms has changed. Having focused on the multidisciplinary deep relationships in nature has changed the simple view of benchmarking them to a scientific and interdisciplinary study. A good example of benchmarking from nature in discussions of non-structural elements in the building is Nano evaluation of the structure of the lotus leaf which has led to the production of self-cleaning glasses (Salsabili et al., 2016). All living creatures in nature have an evolutionary process and change over time based on their needs. In this approach, it has been tried to represent the most fundamental principle of life, which is evolution. Thus, what

eventually occurs is a kind of responsive architecture with the capability of interacting with the environment (Salsabili et al., 2016).

Bionic architecture or the science of studying the living creatures life system has been introduced as one of the three world's supreme sciences (Bionics, Nano, IT). Enlivening buildings is one of the tendencies in bionic architecture. It is a supreme belief that complete coordination or harmony with nature is created (Salsabili et al., 2016). Bionics which is also known as biometrics, or creative life engineering, is the application of systems biology methods of the nature of engineering systems and modern technologies (Salsabili et al., 2016). Bionics, which is also known as biomimicry means to imitate life and originates that is derived from the Greek words *bios* (life) and *mimesis* (imitate). This movement defined and popularized by Janine Benyus in her book "Biomimicry" in 1997. Biomimicry has a small part of the bionic architecture. The contemporary philosophy of architecture is seeking solutions for sustainability in nature by understanding the rules of forms not replicating them (Al-Obaidi et al., 2017). The organic architecture uses nature-inspired geometrical forms in design to reconnect the human with surroundings. Kendrick Bangs Kellogg, an organic architect, believes that "Organic architectures don't take Mother Nature for granted but allow nature to guide their life". So, the form should follow the flow and not work against the dynamic forces of nature. Architect Daniel Liebermann's commentary, an organic architecture, believes that understanding of how we see with our mind and eye is the foundation of everything organic designing. Because our eyes and brains evolved over eons of time and there saw a lot of things. So, we have to go to nature for finding our models now (Pearson, 2001).

2.2.1 History of bionic architecture

Architecture has long drawn from nature as a source of inspiration. Biomorphism, the incorporation of natural existing elements in the design, originated from the beginning of manmade environments and has continued up to now.

The ancient Greeks and Romans incorporated natural motifs into the design, for example, creating a tree-inspired column (Picture 2-1a). Late Antique and Byzantine arabesque tendrils are stylized versions of the acanthus plant. Varro's Aviary at Casinum from 64 BC reconstructed a world in miniature.

The Sagrada Família church by Antoni Gaudí began in 1882 in Spain is a well-known example of using nature's functional forms to answer a structural problem (Picture 2-1b). He used the models of branches of trees as columns to support the vault (Collins, 1963).

Using nature inspiration has long looked but the concept of biomimicry has only started from the middle of the 20th century. René Binet's *Esquisses Décoratives* (1904) predicts and prefigures many of the forms we see today (Picture 2-1c), also a lot of drawings based on biological and morphological was illustrated by Ernst Haeckel's ("Architecture Inspired by Nature: Biomimicry from Art Nouveau to Neo-Futurism," n.d.).

At the 1900 Exposition World Fair in Paris, Binet's Monumental Gate (Picture 2-1d) had been inspired by the most attractive and appropriate representation of the logic and richness of the

tiny shell-like skeletal creatures and he became a precursor of the current trends in parametric architectural design using renderings of natural forms and structures (“Architecture Inspired by Nature: Biomimicry from Art Nouveau to Neo-Futurism,” n.d.).

Pic 2-1: (a) Tree-inspired columns in Greek and Roman buildings, (b) The Sagrada Familia church, (c) Rene Binet's prefigures, (d) Binet's Monumental Gate

Fast forward to the present, Scott Johnson discusses how 21st-century towers are designed by algorithms that are intended to optimize function and performance (Picture 2-2a), and created curvilinear and biomimetic forms.

On temporary architects, design towers inspired not only by biological forms but also by naturally occurring functions (Picture 2-2b) (“Architecture Inspired by Nature: Biomimicry from Art Nouveau to Neo-Futurism,” n.d.).

Pic 2-2: (a) Scott Johnson form, (b) Pearl River Tower in Guangzhou, China

2.2.2 Human interaction with nature

In the way people interact with nature, researchers have proposed this division which is comprised of three distinct perspectives (Salsabili et al., 2016):

- Architecture of nature

The order of nature is more the environment and its related factors such as water, soil, and air. In this architecture, a man tries to change nature with his own desires and overcome it. Unfortunately, the man in this category is in the first place at the outbreak of the negative consequences of environmental degradation, because he has become more dominant on the environment and the planet on which he lives through achieving advanced tools and technology.

In this way, he has made serious damages to the environment and even it can be said that he has disrupted the ecological balance. Of course, this approach will lead to the loss of natural resources. In the past, the scope of this destruction was not too extensive due to a lower population and nature compensating for the gaps by itself. Gradually, nature was no longer able to compensate for these losses due to the increase in the human population and the growth of this destruction.

- Architecture with nature

Consists of the ecosystem, climate and related elements such as the sun, wind and the energies derived from them and its product is sustainable and green architecture. This architecture consists of two sub-categories:

- The attitude to a kind of architecture that aims to have the least damage in the environment and tries to integrate the architecture with nature and these two are not separable. The origin of this architecture can be traced to the era of modern architecture and organic architecture. Examples of such architects were Alvar Aalto and Frank Lloyd Wright.
- In addition to the integration with nature, nature should also be used in a suitable way of making the building self-sufficient in terms of energy and other related issues. This is considered sustainable architecture. In fact, the principle of sustainable architecture should be based on this fact that the building is a small part of the surrounding nature and should act as a part of the ecosystem and its place in life's cycle. Sustainable building goals:
 - Proper utilization of resources and energy.
 - Prevention of air pollution.
 - Compliance with the environment.

- Architecture from nature

It expresses the nature in a broader sense. In this view, everything that man has not intervened in with its creation is called nature. Thus, plants, animals, humans and even cells and perceptions in humans are included. From this perspective, humans try to draw inspiration from nature by looking at its correlation between life and technology, in general, and architecture in particular. Meditation on nature and using appropriate patterns in life is taken from it. This attitude is not separated from the previous attitude, but architecting from nature enables us to reach more sustainability in buildings (Salsabili et al., 2016).

2.2.3 Levels of mimicking in bionic architecture

Biomimicry can work on three levels: the organism, its behaviors, and the ecosystem.

- **On the organism level**, buildings mimic a specific organism without mimicking how the organism with its environment. Because the architects look at the organism itself and try to apply its form and functions.

Norman Foster's Gherkin Tower (2003) has a hexagonal skin inspired by the Venus flower basket sponge (Picture 2-3a, 2-3b). This sponge sits in an underwater environment with strong water currents and its lattice-like exoskeleton and round shape help disperse those stresses on the organism (Ehsaan, n.d.).

On a behavior level, buildings mimic how an organism behaves or relates to its larger context without resistance in its surrounding environment.

The East gate center designed by architect Mick Pearce in Harare, Zimbabwe. The building has no air-conditioning or heating but regulates its temperature with a passive cooling system inspired by the self-cooling mounds of African termites (Picture 2-3c, 2-3d). But its structure does not have to look like a termite mound to function like one and aesthetically made from indigenous Zimbabwean masonry (Fehrenbacher, n.d.).

- **On the ecosystem level**, building mimics the natural process and cycle of the greater environment to work with each other than against. Mimicking the nature of architectural design should follow these six principles in this level:

- Depending on contemporary sunlight.
- Optimizing the system rather than its components.
- Depending on local conditions.
- Diverting in components, relationships, and information.
- Creating conditions favorable to sustained life.
- Adapting and evolving at different levels and rates (El Ahmar, 2011).

The Sahara Forest Project designed by the firm Exploration Architecture is a greenhouse that aims to rely on solar energy alone to operate as a zero-waste system. The project is on the ecosystem level because its many components work together in a cyclical system. The project mimics the Namibian desert beetle (Picture 2-3e) to combat climate change in an arid environment. The greenhouse structure uses saltwater to provide evaporative cooling and humidification. The evaporated air condenses into freshwater allowing the greenhouse to remain heated at night (Picture 2-3f). This system produces more water than the interior plants need so the excess is spewed out for the surrounding plants to grow. Solar power plants work off of the idea that symbiotic relationships are important in nature, collecting sun while providing shade for plants to grow. The project is currently in its pilot phase (Pawlyn, n.d.).

Pic 2-3: Levels of mimicking (a) The Venus Flower, (b) Norman Foster's Gherkin Tower Guangzhou, (c) The mounds of African termites, (d) The East gate Centre in Zimbabwe (e) The Namibian desert beetle, (f) The Sahara Forest Project

2.2.4 Bionic architecture types

Bionic architecture in a general classification can be divided into 6 main categories:

- **Stencils:** this is a cut-out pattern, which has been usually used for decorations and in the Art Nouveau era, its usage peaked. Today, it is used in a functional role, such as front patterns, which are variable based on the amount of sunlight.
- **Structure:** inspiring from the formation of organisms structures and applying them in architecture is possible by understanding the structures and mechanisms forming these living organisms. Inspiration from the structures found in nature requires more research and study than modeling from stencils.
- **Materials:** by studying nature, researchers found that some characteristics of living organisms can be transferred to man-made artificial materials.
- **Form:** in bionic architecture, there is not much value for the pure transference of natural forms to architecture. One important issue to be considered is the relationship between form, structure, and material, as well as the interaction with the environment. Jerry Lebedev suggests that nature is the combination of function, form, and structure, which are effective integrating with each other, and interacting with the environment and adapting to it; otherwise, each alone can do nothing. That is why only translating the shape cannot be considered a

successful experience; however, good architecture is expected with the combination of function and structure.

- **Function:** natural organisms have evolved over hundreds of thousands of years and have been developed in terms of complexity, variety, and versatility. Certain functions that humans learn from different organisms can be changed based on their needs and can be applied in architecture.

- **Process:** inspiring from the processes found in nature that lead to the formation of the phenomenon is more attractive than the form and functional modeling. Perhaps it can be said, this kind of inspiration and impression of nature is the best and most valuable type of inspiration (Salsabili et al., 2016).

2.2.5 The structure of bionic architecture

There are some important factors that exist in nature and humans have utilized them for the construction of buildings including:

- Covering or crust.
- Structure.
- Grooming or decorating.
- Energy.

In general, the designing procedure based on bionics can be stated as follows:

- Selecting a living creature (the given phenomenon in nature):
 - Humans.
 - Animals.
 - Plants.
 - Diatoms, single cells.
- Identifying biological features: life environment, environmental conditions, temperature, humidity, pressure, and sound
 - Reactions: vital resources, respiratory systems, food.
 - Physical characteristics: coexistence conditions, compatibility and incompatibility both direct and indirect.
 - Systematic relationships: the concentration and distribution of bio-statistics, the special geographical conditions.
- Identifying the architectural characteristics
 - The internal structures.
 - Systematic relationships.
 - The main body of the creature: microelements and geometric proportions, macro elements, materials, and proportions (Salsabili, 2014).

2.2.6 Bionic architecture pioneers

- **Leonardo Da Vinci**, the genius person who was the first person in Renaissance that integrated “biologic” and “technology” sciences of his own age to build his flying machine (Picture 2-4a), although he couldn’t success but his trying were the first biometric studies around the world (Sadri, Kavandi, Jozepiri, Teimouri, & Abbasi, 2014; Salsabili, 2014).

- **Greg Lynn** (North Olmsted, OH, 1964), he is popular for his use of animation and motion graphics software for design (computer-aided design). Besides Hani Rashid, Jesse Reiser and, Stan Allen, he was one of the earliest teachers to explore the use of the digital technology for designing building and construction (Picture 2-4b) (Salsabili, 2011). He often experiments with methods of manufacturing from aerospace, boat building, and automobile industries in his designs and his latest works begin to explore how to integrate structure and form together to make some biomorphic forms (Viladas, n.d.).

- **Charlie Luxton**, the pioneer of bionic architecture to create buildings and structures (Picture 2-4c).

- **Paul Klee**, he believed that we should imitate the forces of nature and the world like a child who imitates us in playing games (Picture 2-4d).

Pic 2-4: Bionic Architecture pioneers (a) The flying machine by da vinci, (b) Biomorphc forms by Gerg Lynn, (c) black painted brick house, (d) Paul Klee design

- **Tadao Ando**, he believes that the creation of construction is a part of nature. He had a good penchant for using the figures and configurations as trees, cliffs, mountains, and hills and the renovation and reconstruction of nature of his creative process (Picture 2-5a, 2-5b).

- **Frank Lloyd Wright**, the founder, and theorist of organic architecture and he incorporated his buildings with nature from the very beginning. He believed that it was so hard to distinguish where nature ends and when construction begins (Picture 2-5c).

- **Le Corbusier** had a new look at nature. By designing Villa Savoye in France, he made an intermediate using between inside architecture and outside nature (Picture 2-5d).

- **R. Fisher** believes that today's world is dynamic and based on our needs that constantly change, so the space of living should be dynamic too (Picture 2-5e). For example, if considering our designing ideas and our mood changes, our buildings and constructions have to change according to nature's tune such as shifting from spring to winter, from sunrise to sunset and changing the weather (Salsabili, 2014).

- **Sir Nicholas Grimshaw**, (Hove, 1939) he is a prominent English architect and designed several modernist buildings based on bionic. Including London's Waterloo International railway station (Picture 2-5f) and the Eden Project in Cornwall (Rose, 2007) (Picture 2-6a).

Pic 2-5: Bionic Architecture pioneers (a) Koshino house by Tadao Ando, (b) 4x4 house by Tadao Ando, (c) Falling water house exterior, (d) Villa savoye by Le Corbusier, (e) Rotating building in Dubai, (f) Waterloo station

- **Santiago Calatrava Valls** (Spain, 1951) he is a neo-futuristic architect, structural engineer, sculptor, painter, and etc. He received his bachelor degree in architecture and his master and Ph.D. in civil engineering. Therefore, his designing is a combination of them. Calatrava has tried to bridge the space between structural engineering and architecture. He claims that he continues a tradition of Spanish modernist engineering that included Félix Candela, Antonio Gaudí, and Rafael Guastavino, with a very personal style that derives from numerous studies of the human body and the natural world. Calatrava by knowing sculptor making and painting believes that the practice of architecture is combining all the arts into one (Picture 2-6b). He said, “My goal is always to create something exceptional that enhances cities and enriches the lives of the people who live and work in them” (Picture 2-6c) (“Santiago Calatrava Biography,” 2016).

- **Norman Robert Foster**, (England, 1935), Foster and Partners company are famous for high-tech international design in architecture (Picture 2-6d, 2-6e).

(a)

(b)

(c)

(d)

(e)

Pic 2-6: Bionic Architecture pioneers (a) The Eden Project in Cornwall, (b) El Palau de les Arts Reina Sofia, (c) The City of Arts and Sciences, (d) High tech Buildings, (e) Millau Bridge

- **Ken Yeang** (Malaysia, 1948), he is an architect, ecologist, author and he is famous for his signature of eco-architecture and eco master plans. He started ecology-based green design since 1971 and Guardian named him "one of the 50 people who could save the planet". Yeang's early work is creating low-energy passive-mode buildings.

In the 1990s, Yeang started working on designing the high-rise typology as "Vertical green urbanism" or "city-in-the-sky". So, he is the inventor of the "Bioclimatic skyscraper" by completing over 12 bioclimatic eco high-rise buildings (Picture 2-7a). It continues in 2011, his ideas of making a park in high-rise buildings are manifested as a "Vertical linear park". This type of park led to his concept of "Green eco-infrastructure", a device that makes a vital ecological nexus between the built form and its surrounding landscape (The iconic SOMA

Masterplan in Bangalore, India, Picture 2-7b) and (The Spire Edge Tower in Gurgaon, India, completion c. 2013, Picture 2-7c). Then, this green eco-infrastructure concept helps him to develop it as a "eco-master planning" that is the weaving of four eco-infrastructures to create a unified system (Hart, 2011).

- **Daniel Libeskind** (Poland, 1946), he is an architect, artist, professor and set designer (Picture 2-7d) (Libeskind, 2004).

- **Jan Kaplický** (1937 –2009), he was a world-renowned neo-futuristic. In 1979, he set up his own architectural think with David Nixon called Future Systems and developing an architectural style of combining organic forms with high tech futurism (Picture 2-7e). He believes that the world is full of beautiful things and we have to be observant as an architect otherwise we are in trouble. He also believes that creativity is everywhere so we don't need to go anywhere particular to be creative, creativity depends on our relationships with other people, our personal relationships, our partner or whatever, and creativity depends on our moods too, our moods (happiness or unhappiness) comes out in our work and reflects in our emotional state and we can't separate the two (Mikhail, 2003).

- **Cecil Balmond**, His method of designing is derived from theories of complexity, non-linear organization, and emergence. He investigates mathematical concepts and their influence on

Pic 2-7: Bionic Architecture pioneers (a) Bio climate high rise building, (b) The iconic SOMA Master plan in Bangalore, India, (c) Westside Shopping and Leisure Centre, (d) The Selfridges Building, England, (e) Czech national library, (f) Balmond studio

natural forms and structures, interrogating algorithms, fractals, rhythm and cellular structure (Picture 2-7f) (Linden, n.d.).

2.2.7 The most popular bionic examples in architecture

- The Selfridges Building (Birmingham, England, 2003)

It designed by architecture Jan Kaplický and Amanda Levete group. The building is part of the Bullring shopping center and houses Selfridges department store. It has a steel framework with the diffuse concrete facade. From its construction, it has become an iconic architectural landmark of Birmingham (Picture 2-8a).

- Olympiastadion (Munich)

This included large sweeping canopies of acrylic glass stabilized by steel cables that were used for the first time on a large scale (Picture 2-8b).

- Turning Torso

It is a white marble piece based on the form of a twisting human being was designed by Calatrava (Picture 2-8c). This immobile building is solid and the first twisting tower. This building consists of nine segments of five-story pentagons that twist relative to each other around the vertical core as it rises and it is supported by an exterior steel framework.

- Oita Bank Dome

Kisho Kurokawa designed this stadium. It has a retractable dome roof with a wire traction system. (Picture 2-8d)

Pic 2-8: The most bionic architecture examples (a) The Selfridges Building, (b) Olympiastadion, Munich, (c) Turning Torso, (d) Oita Bank Dome

- The Bionic Tower

It is a vertical city, it designed by Eloy Celaya, María Rosa Cervera, and Javier Gómez. It is eco-friendly and after completion, it will become the tallest building in the world. This bionic tower is made of twelve vertical neighborhoods with eighty meters high of each of them and the base island with 1,000 meters in diameter. Each neighborhood is separated to make for easier construction and depletion in case of an emergency and safety (Picture 2-9a). Besides, the base of Iceland made up of many buildings, pools, gardens, and communication infrastructures.

- The Dynamic Tower (Da Vinci Tower)

It was proposed by 420-metre with a capacity of rotating floors independently and changing the shape of the tower (Picture 2-9b). Each floor is designed to rotate a maximum of 6 meters (20 ft.) per minute or one full rotation in 180 minutes. The entire tower is powered by wind turbines located between each of the rotating floors and solar panels. Its surplus electricity could be used to power five other similar sized buildings in the vicinity.

- City Hall, London

It is located on the River Thames in London. The building has an unusual, bulbous shape (Picture 2-9c), to reduce its surface area and improve energy efficiency, but it has a double facade with glass exterior facade overwhelms the benefit of shape. So, cause it to be an inefficient building in terms of energy use (375 kWh/m²/yr.), and getting a rating of “E” in 2012.

- The Ascent at Roebling's Bridge

Daniel Libeskind designed a residential building with 70 condominiums. Libeskind cites the Ohio River and the Roebling Bridge as influences for his design. The concrete structure slopes outward from its base on its eastern face and is clad in a glass curtain wall. (Picture 2-9d)

Pic 2-9: The most bionic architecture examples (a) The Bionic Tower, (b) The Dynamic Tower, (c) City Hall, London, (d) The Ascent at Roebling's Bridge, (e) The City of Arts and Sciences

- The City of Arts and Sciences

It is an entertainment-based cultural and architectural complex in the city of Valencia, Spain. It is the most important modern tourist destination in the city of Valencia and one of the 12 treasures of Spain (Picture 2-9e). The complex is made up of the following buildings:

- L'Hemisfèric, The building seems like a giant eye. The hemispheric was also known as the planetarium or the “eye of knowledge” (Picture 2-10a). It was the first building completed in 1998 in the center of the City of Arts and Sciences. This planetarium is a half-sphere composed of concrete over the pool. The bottom of the pool is glass. The

shutter is made of aluminum awnings that fold upward collectively to form a roof that opens along the curved axis of the eye. There is a miraculous echo inside of the building and if two people stay on the two opposite pillars inside of the eye they can seamlessly speak with each other.

- El Museu de les Ciències Príncipe Felipe (2000), it is a museum of science that resembles the skeleton of a whale. The building's architecture is known for its geometry, structure, use of materials, and its design around nature. The building is currently the largest place in Spain (Picture 2-10b).

Pic 2-10: The most bionic architecture examples (a) L'Hemisfèric, (b) El Museu de les Ciències Príncipe Felipe, (c) L'Umbracle, (d) L'Oceanogràfic buildings, (e) El Palau de les Arts Reina, (f) El Pont de l'Assut de l'Or, (g) L'Àgora

- L'Umbracle (2001), an open structure walk with species plants in Valencia. The plants change their color with each season to make a variable place. It designed by Miquel de Navarre, Francesc Abbot, Yoko Ono and others (Picture 2-10c).

- L'Oceanogràfic (2003), Félix Candela designed this open-air oceanographic park. This aquarium is home of over 500 different species including belugas, sawfish, dolphins, starfish, jellyfish, sea urchins, walruses, sea lions, seals, penguins, turtles, sharks, rays, and wetland bird species (Picture 2-10d).

- El Palau de Les Arts Reina Sofia (2005), it is an opera house and arts center. It contains four large rooms: the main room, amphitheater, magisterial classroom and theater of

camera. The building is supported by white concrete with two laminated steel shells to cover (Picture 2-10e).

- El Pont de l'Assut de l'Or (2008), it is a bridge with a white cable-stayed crossing the dry Turia riverbed. Its tower has 125 meters high and was placed at the highest point in the city (Picture 2-10f).

- L'Àgora (2009), The Agora is a space designed to hold a variety of events such as concerts, performances, exhibitions, conventions, staging of congresses, and international sports meetings (Picture 2-10g).

- The Beijing National Aquatics Center

It is a water center that was built for the swimming competitions of the 2008 summer Olympics (Picture 2-11a). It made of steel space frame and it is the largest ETFE dressed structure in the world with over 100,000 m² of ETFE pillows that are only 0.2 mm in total thickness, these pillows are lighter and more heat permeation than traditional glass and make a 30% decrease in energy costs. The outer wall is based on the structure devised from the natural pattern of bubbles in soap lather and it is really a rectangular box. The building's popularity has made many copies if it in China. For example the facade near the ferry terminal in Macau or the Casino Oceanus by Paul Steelman.

- The 'Roof-Roof' House (1985)

Yeang designed his house in Kuala Lumpur, it was his first experimental of bioclimatic built work. He brings several experimental ideas with a single one curved louvered. It rolls like an umbrella to filter environment solar, shading solar-filtering device and a second shading the terrace (Picture 2-11b). The large louvers are angled to let in the easterly morning sun but keep out the hot mid-day and western sun. It has side "wind wing-walls" at the south to direct wind into the dining area. On the east is a pool for swimming also it functions as an evaporative-cooling device to cool easterly breeze before entering internal living spaces (Picture 2-11c). The influences of this building form bioclimatic ideas for later works of Yeang (Hart, 2011).

- The Mesiniaga Tower (IBM Franchise)

This tower brings various earlier experimental "bioclimatic skyscraper" ideas in a single built form. It has these benefits (Picture 2-11d):

- The placement of the elevator core as a solar shield at the tower's hot side
- The placement of toilets and stairwells to receive natural ventilation opportunities
- The various solar-path shaped sun-shades
- The use of an evaporative-cooling pool at the uppermost roof level
- The overhead louvered canopy as a framework for future PV cells
- The vegetated and stepped facade (Picture 2-11e).

Yeang received several awards including the Aga Khan Award for the architecture of this building (1993). His ideas invert the high-rise typology to a “city-in-the-sky” or “vertical urban design”, which he first examined in his high-rise National Library in Singapore (2005) (Hart, 2011).

Pic 2-11: The most bionic architecture examples (a) The Beijing National Aquatics Center, (b) The 'Roof-Roof' House, (c) The 'Roof-Roof' House diagrams, (d) The Mesiniaga Tower, (e) The Mesiniaga Tower diagrams

2.2.8 The most popular bionic examples in other fields

- **Water Turbines was modeled from Humpback Whales**, the bumps found that are located on the fins of a humpback whale made the inspiration of the wind turbine blades. Designing based on it has made the turbines quieter, more reliable and even better in turbulent winds (Picture 2-12a, 2-12b) (Seth, 2009).

- **Water Mixer by Pax Technologies was modeled from Lily**, The centripetal spirals of the lily flower inspired designers to make water mixer with the shape of flowing of liquid (Picture 2-12c, 2-12d).

- **Bullet Train was modeled based on The Kingfisher**, for resolving the extremely loud claps that occurred when train emerged from tunnels, Japanese engineers inspired a Kingfisher Bird and how effortlessly this bird dove into the water with his beak to make a new train (Picture 2-12e, 2-12f).

Pic 2-12: The most bionic examples (a) Water Turbines, (b) Humpback Whales, (c) Water Mixer, (d) Lily flower, (e) Bullet Train, (f) The Kingfisher

- **Bionic Penguins**, these types of creature is a set of Bionic Penguins inspired by penguins that can navigate through the seas for operating and searching (Picture 2-13a).

- **Air_ray by Festo**, This balloon floater takes its inspiration from the manta ray. It's a remote-controlled hybrid floater with a helium-filled ballonnet that could drive (Picture 2-13b).

- **Lotusan Paint was modeled based on Sacred Lotus**, The complex surface of large-winged insects as butterflies and lotus have hydrophobic surfaces. This type of surface allows self-cleaning because of their exterior coating. This coating uses the principles of the microstructure to automatically become clean after a rain-shower (Picture 2-13c, 2-13d).

- **Morphotex Structural Colored Fibers was modeled based on Morpho Butterflies**, Butterflies' wings are made of many layers of proteins that refract light in different ways. So, Morphotex fibers were inspired by these vibrant blue of the butterflies (Picture 2-13e, 2-13f).

Pic 2-13: The most bionic examples (a) Bionic Penguins, (b) Air_ray, (c) Sacred Lotus, (d) Lotusan Paint, (e) Morpho Butterflies, (f) Morphotex Structural Colored Fibers

- **Philips Webcam by Flynn Product Design**, creepy animals evoke creepy designs as well, like this Philips webcam by Flynn Product Design. Philips webcam is created based on spider and octopus legs; their legs help re-position the camera to a favored nook (Picture 2-14a).

- **Lunocet Monofin**, fish-finis an ideal aid for swimmers. It's made from silicon covered in carbon fiber with a titanium footplate. This system gives you the same power in downstroke and upstroke (Picture 2-14b).

- **Mercedes-Benz was modeled based on Boxfish**, the form of boxfish made the base of 2005 Mercedes-Benz. This very cute and simplified car has a 65% lower coefficient than other compact cars of that time (Picture 2-14c, 2-14d).

2.3 Nature

Nature is an unlimited source of inspiration and a good teacher in various fields of knowledge, technology, science and architectural design (Taghizadeh & Bastanfard, 2012). Because nature is God-like, realizing life through the manifestation of "The Spirit" (Case, Wright, & Ellis, 1986). The systems, which are available in nature, is the tendency of enlivening the building in bionic architecture. Charlie Luxton the pioneers of Bionic architecture believes that the main result of this architecture is the best using of materials to make the strength of building and architectural flexibility and variety. So, we have to focus on the rules of laws that govern nature (Taghizadeh & Bastanfard, 2012).

Pic 2-14: The most bionic examples (a) Philips Webcam, (b) Lunocet Monofin, (c) Mercedes-Benz's, (d) Boxfish

- History of human interaction and nature

Human history has been contacting with nature. This relationship divided into the following four types based on the forms of relations with nature:

- Nature and organic model: the time of Hunting Age or Stone Age, a human being has been overcome by nature in all positions of life.
- Nature and Hyper organic model: the farm age, trying to be comforted by nature of having security, happiness and with the growing up the sense of being "With nature".
- Nature and dominate nature model: the industrial revolution with the aspect of security, justice, and welfare made the maximum utilization of human nature.
- Nature and industrial quasi-organic model. Nowadays, Despite being the motivation of using natural, but the global community is trying to adopt industry to nature (Picture 2-15) (Taghizadeh & Bastanfard, 2012).

Pic 2-15: The four steps of relation between human and nature

2.4 The human body

Each natural element is the complex of several components and elements to make a system. Studying nature enable us to achieve these systems. Each of these natural systems is consist of a number of subsystems, and each of them is also working as a separate system. For example, regarding function, the human body is consists of a respiratory system, digestive system, visual system, auditory system, etc. Each of them has an interaction with the other systems besides, they are formed of several components too (Taghizadeh & Bastanfard, 2012).

2.4.1 History of the human body and architecture

The standard for the human body is not born in the 20th century but it belongs to the Renaissance.

- In 1487, Leonardo da Vinci drew "The Vitruvian Man" (Figure 2-1a) the most famous drawings of human body scales and it becomes the main sort of addressing the question of the standard since then ("How do the musculoskeletal and cardiorespiratory systems of the body influence and respond to movement?," n.d.).
- In 1936, Ernst Neufert, one of the first student of Bauhaus and Walter Gropius, creates an Architects Databook. This book consists of a relation of the human male body and its environment (furniture, street, building etc.) (Figure 2-1b). His dimensions diagrams have been considered as a source of fundamental information in many projects since then. This

book is still considered as the “Bible” in some countries such as Brazil (“How do the musculoskeletal and cardiorespiratory systems of the body influence and respond to movement?,” n.d.).

- The good example of the proportional of the human body is, Le Corbusier’s modular which was supposed to support the range of measurements of the human scale, despite working hard on this models, this model just contains man scales (Figure 2-1c) (“How do the musculoskeletal and cardiorespiratory systems of the body influence and respond to movement?,” n.d.).

- In the second part of the 20th century, similar diagrams were collected as the Architectural Graphic Standards. It used as a reference by many architecture offices because of containing a dimensioned combination of the body and architecture (Figure 2-1d) (“How do the musculoskeletal and cardiorespiratory systems of the body influence and respond to movement?,” n.d.).

Fig 2-1: The history of human body and architecture (a) The Vitruvian, (b) Neufert data book, (c) Le Corbusier’s modular, (d) Architectural Graphic Standards, (e) Henry Dreyfuss standards, (f) Philippe Rahm and Catherine Mosbach standards

- In 1974, American industrial designer Henry Dreyfuss re-interprets these diagrams by adding the gender (by the names of Joe and Josephine), wheelchair users, handicapped and elderly person to them (Figure 2-1e). Differentiation as well as the acknowledgment that other bodies exist and respond differently to normalized environments (“How do the musculoskeletal and cardiorespiratory systems of the body influence and respond to movement?,” n.d.).

- In the recent year (2011), Philippe Rahm architects and Catherine Mosbach made a new diagram of the human body in relation with its environment based on anatomical dimensions

and biological affections by the environment such as humidity, heat, and pollution. These three factors are having physiological consequences on the body so they are mapped and exploited in the creation of proposition (Figure 2-1f) (“How do the musculoskeletal and cardiorespiratory systems of the body influence and respond to movement?,” n.d.).

The most difficult stage of mastering the natural forms in architecture is a period from the middle of 19th until the beginning of the 20th century because of the rapid development of biology and building technique by the invention of steel concrete and using of glass and metal in structures (Zakharchuk, 2012).

2.4.2 The human body and design of buildings

From the beginning of the human life, having relation with nature solve the challenges of making building with its details such as mechanical and electrical, structural, protection, cooling, shell building and security and so on. Besides, bionics solve the gap between nature and technology. Nature involves sustainable development of complex systems. Furthermore, bionics is knowledge of a combination of different concepts of natural sciences, formed of complex biological systems and living organisms and engineering to solve the technical problems of structural engineering. So, bionics structures caused a revolution in the development of technology and it is still being developed (Taghizadeh & Bastanfard, 2012).

2.4.3 The human body, model for future structural design

Architecture and engineering have traditionally to treat structurally as static and to resist all anticipated loads. The human body as a model suggests a lighter and dynamic structural system that actively responds to its forces and controls their position to handle resonant motion and building drift. So, the human body is a good example of building for the near-future buildings because of its responsibility for efficiently and dynamically behavior that has to force the load placed upon them (Taghizadeh & Bastanfard, 2012). Besides the body shape, the human specter can be re-shaped too. So we have to work with ordinary materials and turn them into something else to make new structures (“Lucy McRae’s Body Architecture,” 2010).

2.4.4 Structure of the human skeleton

Based on biomechanics, the human body is a complex and variable set of levers to balance the body against the external forces by using muscles around the joint and having brain and nervous system to control and sending a message to muscles.

2.4.4.1 The human joints

Joints are formed at the junction of two or more bones. Different types of joints provides varying degrees of movement. The joints divided into rigid and movable joints. In rigid joints (skull joints) because of excessive force, we have stability at the junction, but moveable joints divided to semi-movable joints (spine joints) with stability and somewhat flexible and synovial joints (hip joint) can move freely. Thus detailed of movable joint connection pattern,

can be a good model for the future architectural construction because of their mobile and flexible structures (Taghizadeh & Bastanfard, 2012).

2.4.4.2 The human brain and the nervous system

The nervous system has a special structure to coordinate actions between differentiated cells and organs of the body and it was evolved during these previous years. Nervous system reacts in two main categories of stimulating and sending commands to other components. The nervous system is divided into two groups based on the cognitive framework:

- The central nervous system includes the brain and spinal cord.
- 12 pairs of nerve fibers in the brain and 31 pairs of nerve cord that connect the brain and spinal cord to other parts of the body (Taghizadeh & Bastanfard, 2012).

Due to the abilities of the nervous system, the human brain is a good model for a strong building because it could be:

- Response to any change in the structural connections.
- Response to movement building in earthquake, wind.
- Response to vandalism and (in case of military attack and etc.).
- Response to electrical system problems.
- Response to mechanical problems (water, sewer and etc.).
- Response to changes in humidity and temperature inside the building.
- Response to smoke in indoor (Taghizadeh & Bastanfard, 2012).

2.4.4.3 The human muscles

Each human body has 640 muscles in different shapes and sizes. They make up 40 percent of our human bodies weight. Muscles in groups provide us stability and ability to do many functions such as moving, breathing, running, walking, writing and dancing. There are three types of muscles in the body:

- Cardiac: an involuntary muscle that is only found in the heart. These uncontrolled muscles are striated.
- Smooth: these involuntary muscles with long and slim fibers are non-striated. They are found in the walls of organs such as the stomach, bladder, arteries, and veins. They Contract slowly and sustain.
- Skeletal: these voluntary muscles are the most common types of muscles in the body. They are attached to the bones and named according to their locations. They are controllable, striated in texture, contract easily, require rest after exercise, have tremendous power and strength (“How do the musculoskeletal and cardiorespiratory systems of the body influence and respond to movement?,” n.d.).

2.4.5 The human body as a solution to building problems

The biomimetic architecture looks to nature as a model, measure and tutor to solve buildings problems in architecture, modeling, imitating and taking inspiration from natural designs and processes to provide their man-made designs.

In the past, the biomorphic architecture used natural existing elements just as a source of inspiration for aesthetic components or forms (Picture 2-16a) (Benyus, 1997). So according to what Yeang said, over 4 decades ago humans are using nature and environment unfeeling and affecting adversely the planet's natural balance (causing global warming and climate changing) (Hart, 2011).

In a technological age, architectural innovations can mimic natural forms, functions, and processes (Picture 2-16b) by responding to prior architectural movements of increasing the efficiency of resources (rely on solar energy instead of fossil fuels).

Today, bionic architectures use a wide range of new methods by using nature in their designing such as mathematical algorithms, genetic algorithms, neural networks, algebra fractal and algebra disturbed (Cheras) (Picture 2-16c). Their designs suggest that analyzing mathematical methods of the system also follow natural patterns (Ghafoury Pour, Shemirani, & Sheikh, 1393).

Pic 2-16: (a) Using the nature unfeeling, (b) Natural form, (c) Bionic architecture by combined of mathematical method

There are extensive examples of nature for modeling and many of them have already been used. In the meantime, but taking the advantages of the human body especially in the field of construction has been neglected. At first glance, it might be thought that the human body is very complex, but having studied this point, we can conclude that in spite of our imagination, it is much understandable and usable structure. Also, there have been many studies about its behavior.

2.5 Preliminary conclusions

According to what has been presented in this chapter, an architect acquires a new degree of freedom to create structures based on the recent advances in materials, construction techniques, calculation methods, and structures to make beneficial structures based on nature. Their fields of use can be categorized (Taghizadeh & Bastanfard, 2012):

- Using the most adequate shapes

Shapes in nature are formed for the best function and using the least material. As Timpson said the living creature shapes are the result of matching with physical powers and their wise reaction for a living. In fact, form and structure are the results of compromising with the environmental powers (Ghafoury Pour et al., 1393).

- Using features and characters

Nowadays, the modern stage of learning shapes of nature in designing are formed based on the laws of wild nature and architecture. Nowadays architects besides using the outer forms of wild nature use the features and characteristics of a form (Zakharchuk, 2012).

- Using aesthetic form

The structural forms of nature can be used in different types of buildings such as large-span and high-rise buildings or quickly transforming constructions. Architectural bionic not only decides functional questions of architecture but also opens the good outlook in searching of a function and an aesthetic form of architecture combination, and finally, it teaches the architects to think how the using of synthetic forms and systems (Zakharchuk, 2012).

- Using functional and comfort form

A bionic style by the combination of natural forms and high tech and as the fastness of modern life creates the bright standards for the vital area, one of the most important standards of modern housing quality by bionic style is its functionality and comfort.

- Using the dynamic form

At all times, the architecture character was mixed of organic by the laws of nature and an artificial by their ideals. But the organic architecture designed the forms not only the geometry of nature but also based on dynamic, irregular and a result of nature contact with reality. So their designs are right with perfect proportions, golden section and the absolute beauty with regarding the result of education, and the human body is a good example of nature to simulate (Zakharchuk, 2012).

- Using form in motion

One of the features of nature is its ability to move and change. Basically one of the points of small regularity in-universe is a movement like changing place or shape. Significant note in structure movements like the human body is the balance while moving.

Finally, using the architecture of human beings has been extended and deepened in recent years to make popular agreeable space by human consciousness (Newman, 1972).

CHAPTER III: THE SKELETAL SYSTEM

3.1 The skeletal system

This section will present the findings from the study of the skeletal system. The skeletal system is the framework of the body and it is composed bones, cartilage, synovial fluid, discs and ligaments which are compacted to each other in human joints.

3.1.1 Cartilage

Articular cartilage is covered by our joint surfaces. Normally, this is a smooth well-lubricated surface whose frictional resistance is less than that of an ice skate gliding on ice (Picture 3-1) (Frederick A. Matsen III, 2011).

It does not have a blood supply. Even it gets its oxygen and nutrients from the surrounding joint fluid. In the loading situation, the fluid including waste products squeezes out of the cartilage because of pressure, the fluid seeps back into oxygen and nutrients after relieving the pressure. So, the health of cartilage depends on its using and when it is injured, it has a limited ability to repair itself (Frederick A. Matsen III, 2011).

- **Character:** Normal cartilage is very stable, slightly elastic, and a shock absorber of our joints.

Pic 3-1: Cartilage place

3.1.2 Synovial fluid

A thin film of synovial fluid covers the surfaces of synovial membrane and cartilage in the joint space in the human body (Picture 3-2) (Frederick A. Matsen III, 2011).

- **Character:** Synovial fluid is the same as an adhesive seal that allows sliding motion freely between cartilaginous surfaces even in small articulations (Frederick A. Matsen III, 2011).

Pic 3-2: Synovial place

3.1.3 Ligaments

They are hard fibers of connected tissues. In the skeletal system ligaments connect one bone to another bone, tendons connect muscle to bone, and fasciae connect muscles to other muscles (Picture 3-3) (Dorland, 2007).

- **Character:** ligaments have several functions as follow:

- Protection: during the moving, ligaments protect the bones from contacting each other. Besides, they absorb the stress upon the skeleton.
- Dictating the Range Of Motion: the ligaments keep all the structures together and control the range of movements.
- Proprioception: ligaments have the ability to know the position of each part of human bodies and participating in the proprioception (Cucic, 2017).

They have various shapes and two types:

- White ligament: they consist of collagenous fibers so they are strong and inelastic.
- Yellow ligament: they consist of elastic fibers so they are quite hard and elastic (Britannica, 2007).

Pic 3-3: ligaments

3.1.4 Discs

Disks are shock absorber between each of the separated vertebrae in the spine structure and protect the nerves in the middle of the spine and intervertebral disks (Picture 3-4).

Pic 3-4: Disc

- **Character:** We have 24 disks between vertebrae. These are made of fibrocartilaginous material. The outside of them have strong protective covering material (the annulus fibrosus) and the inside have a jelly-like substance (the nucleus pulposus) to get a better position to absorb pressure (“Intervertebral Disc,” n.d.).

3.1.5 Bones

Our body is shaped with 206 bones (20 percent of our body weight) and it enables us to have function each day.

3.1.5.1 Bones Character

- **Protection:** vital organs are protected with the bones of the body. For example, the pelvis surrounds the reproductive organs.
- **Movement:** bones have the same role as levers to allow the body to move in conjunction with muscles. For example, the flexion of the knee joint allows us to kick a ball.
- **Formation of blood cells:** the cavities in certain bones make the formation of blood cells.
- **Support:** bones are the framework of the body. So, they support our body and give it shape. For example, the vertebrae support the ribs to enable us to stand.
- **Storage of minerals:** bones is the store of minerals, which are needed for the functioning of the body. For example, calcium is stored in bones (“How do the musculoskeletal and cardiorespiratory systems of the body influence and respond to movement?,” n.d.) (Picture 3-5).

Pic 3-5: Major bones involved in movement

3.1.5.2 Bone Types

Bones are categorized into five different types, as follow:

- Flat bones: these bones have the role of protecting the internal organs. Such as scapula (shoulder blade), the pelvic girdle, the skull (cranium), sternum (breastbone) and ribs.
- Sesamoid bones: these are small bones that they are held within tendons. Such as the patella (knee cap).
- Long bones: these are the bones, which are connected to do large movements. They are long, cylindrical and have growth heads in epiphysis. These bones have the hard part “compact bone”, the outer layer of the bone, and spongy part “cancellous bone” the inside of the bone. Such as long bones include the femur (thigh bone), the humerus (upper bone in the arm) and the phalanges (fingers and toes).
- Short bones: these bones, which are small with complex movements, are almost in cube shapes. Such as the carpals (small bones in the base of the hand) and tarsals (in the feet).
- Irregular bones: the shapes of these bones are irregular. Such as the vertebrae and some facial bones (“How do the musculoskeletal and cardiorespiratory systems of the body influence and respond to movement?,” n.d.).

3.1.5.3 Bones structures

Bones structures are light enough to make movements and strong to support our body weight.

The bones contain (Picture 3-6):

Pic 3-6: Bones structure

- At first, a thin layer of dense connective tissue known as Periosteum. The periosteum provides a good blood supply to the bone and a point of muscular attachment.
- Under the periosteum, there is a thin layer of compact bone known as cortical bone, which provides the bones strength.
- On the inside of both end of the bones, there is spongy bone (cancellous bone). This is porous, lightweight and having an irregular arrangement of tissue interestingly.
- The center and hollow part of the bone shaft is the Medullary Cavity. This is contained in yellow bone marrow (a fatty tissue) and the red bone marrow (blood cells) (“Structure of Bone,” n.d.).

A long bone consists of these parts: Diaphysis (This is the long central part), Epiphysis (The rounded ends of long bones), Metaphysis (Area between the diaphysis and epiphysis at both ends of the bone), and Epiphyseal Plates (these are Plates of cartilages (growth plates) covering the ends of bones during childhood then replaced by bone after that times).

3.2 Joints

Joints (joined) are the past participle of the English word of the join. It is derived from Latin iunctus, past participle of the Latin verb iungere, join together, unite, connect, and attach (Lewis, Freund, & Short, 1969).

A joint or articulation (or articular surface) is a functional package that they are formed to allow for different movement and degree with their connections between bones in the body (Whiting & Rugg, 2006).

Bones are joined in the human body to each other to bring a variety of ways to serve the functional requirements and purposeful motion of the musculoskeletal system. Muscles, tendons, ligaments, cartilage, and bone all do their individual share to create different function. The normal joints have two important characteristics of stability and lubrication (Frederick A. Matsen III, 2011).

3.2.1 Types of human joints

Holding the skeleton together and supporting the movement are the roles of joints in our bodies. There are seven types of categorizing joints but two of them are main. The first is joint types by function and motion and the second is joint types by material and structure.

3.2.1.1 Joint types by their function and motion

Based on their function and motion. The joints divided to:

- Immovable joints (Synarthroses) such as skull sutures, the articulations between the teeth and the mandible, and the joint between the first pair of ribs and the sternum.
- Slight movement (Amphiarthroses) are included the distal joint between the tibia and the fibula and the pubic symphysis of the pelvic girdle.

- Full movement (Diarthrosis) are included in many bone articulations in the upper and lower limbs. Examples of these include the elbow, shoulder, and ankle (Picture 3-7) (Table 3-1) (“Bones Come Together: Types of Joints in the Human Body,” n.d.).

Pic 3-7: Joint types By Their Function and Motion examples

Table 3-1: Types of Joint function

Type of Joint Function	Examples
Synarthrosis (no movement)	Skull Sutures, articulations of bony sockets and teeth in the facial skeleton (e.g. sternocostal, tibiofibular)
Amphiarthrosis (little Movement)	The distal joint between the tibia and the fibula and the pubic symphysis (e.g. vertebral bodies, pubic bones)
Diarthrosis (full movement)	Elbow, shoulder, ankle (most appendicular)

3.2.1.2 Joint types by their structure

Due to their structures, the joints divided to three subset:

- Fibrous joints (Synarthrodial): ligaments are the only material to make this type of joints. For examples, where the teeth are held to their bony sockets and at both the radioulnar and tibiofibular joints.
- Cartilaginous (Synchondroses and symphyses): cartilages are the only material that is used to connect between articulating this type of joints. The connection between vertebrae in the spine is the best example of this type.
- Synovial Joints (Diarthrosis): They are highly moveable and the most common classification of the joint in the human body. These joints made up the synovial capsule (collagenous structure) surrounding the entire joint. This group is categorized into six

types by the shape of the joint and the movement (“How do the musculoskeletal and cardiorespiratory systems of the body influence and respond to movement?,” n.d.).

3.2.1.2.1 Fibrous joints

Most of these type of joints are immovable because of thick connective tissues between the articulations. Fibrous joints categorized into three types such as:

- Gomphoses: the immovable fibrous articulations between the teeth and the mandible or maxilla are the example of these joints.
- Sutures: immovable joints that connect bones of the skull. These joints contain serrated edges, which are locked together with fibers of connective tissue.
- Syndesmosis: a little movement joint, which is connected by two bones with a ligament. A good example of this joint is the distal joint between the tibia and fibula (Picture 3-8a) (“Bones Come Together: Types of Joints in the Human Body,” n.d.).

Pic 3-8: Joints type by their structure (a) Fibrous Joints examples, (b) Cartilaginous Joints examples, (c) A cartilaginous joint between two vertebrae

3.2.1.2.2 Cartilaginous joints

In these joints, bones are connected to each other with cartilage. These joints divided into two types:

- Synchronous: this type is temporary and immovable joints, which are only present in children, up until the end of puberty. For example the epiphyseal plates in long bones and the joint between the first pair of ribs and the sternum (“Types of Joints,” n.d.).

- Symphysis: this permanent joint (“Types of Joints,” n.d.) with some movements consists of a compressible fibrocartilaginous pad between two bones. This type of joint allows for some movement. The examples of this joints are the hip bones which are connected by the pubic symphysis, and the vertebrae with intervertebral discs (Picture 3-8b, 3-8c) (“Bones Come Together: Types of Joints in the Human Body,” n.d.).

3.2.1.2.3 Synovial joints

These movable joints in the body of mammals are characterized by having an articular capsule between the two joined bones and protecting by a coating of articular cartilage, and its surrounding ligaments help joints to be supported and reinforced in limit movement (Drake, Vogl, & Mitchell, 2009). These joints are categorized into six types (Picture 3-9) (Table 3-2):

- Hinge: these joints allow for extension and flexion on just one axis. It includes the elbow, knee, and finger joints.
- A saddle: this joint allows for extension, flexion, and other movements, but no rotation. It includes the thumb’s saddle joint in the hand (between the first metacarpal and the trapezium).
- The ball-and-socket: this joint has a freely moving and rotating on each axis. It includes the hip and shoulder joints (“Bones Come Together: Types of Joints in the Human Body,” n.d.).
- Gliding: this joint is moving against each other on a single plane. It includes the bones of the wrists, ankles, and intervertebral joints.
- A pivot: this joint provides rotation. It includes at the top of the spine allows for rotation of the head.
- A condyloid: this joint allows for circular motion, extension, and flexion. It includes the wrist joint between the radius and the carpal bones.

Pic 3-9: Synovial Joints examples

Table 3-2: Synovial joints types

Joint Type	Movement at joint	Examples	Structure
Hinge joints	These joints act as a door, allowing flexion and extension in just one plane (Flexion/Extension)	Elbow/Knee
Pivot joints	Rotation of one bone around another	Top of the neck
Ball and socket joints (universal Joint)	These allow for all movements except gliding (Flexion/Extension/Adduction/Abduction/Internal & External Rotation)	Shoulder/Hip
Saddle joints	which resemble a saddle, permit the same movements as the condyloid joints but allow greater movement (Flexion/Extension/Adduction/Abduction/Circumduction)	CMC joint of the thumb
Condyloid joints (ellipsoidal joints)	A condyloid joint is where two bones with odd shapes of concave and convex fit together. (Flexion/Extension/Adduction/Abduction/Circumduction)	Wrist/MCP/MTP joints

Plane joints (gliding joint)	These joints allow only gliding or sliding movements, are multi-axial such as the articulation between vertebrae. (Gliding movements)	carpals of the wrist/ acromioclavicular joint
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3.2.1.3 Joint types based on axes of movement

A number of axes of movement are the main character of classifying:

- Nonaxial.
- Monoaxial.
- Biaxial.
- Multiaxial (Cucic, 2017).

3.2.1.4 Joint types based on degrees of freedom

Another classification is based on the degrees of freedom that they are allowed, and categorized to one, two or three degrees of freedom (Cucic, 2017).

3.2.1.5 Joint types based on number and shapes of the articular surfaces

This classification is according to the shapes and number of the articular surfaces. It categorized to flat, concave and convex surfaces (Platzer, 2004).

3.2.1.6 Joint types based on the biomechanical classification

Joints can also be classified based on their number of bones involved. So they are divided into simple, compound and complex depending on the number of bones involved.

- Simple joint: two articulation surfaces (e.g. shoulder joint, hip joint).
- Compound joint: three or more articulation surfaces (e.g. radiocarpal joint).
- Complex joint: two or more articulation surfaces with an articular disc or meniscus (e.g. knee joint) (Johnson, n.d.).

3.2.1.7 Joint types based on the anatomical classification

The joints are classified anatomically into some groups as follow:

- Elbow joints.
- Stern clavicular joints.

- Vertebral articulations.
- Sacroiliac joints.
- Hip Joints.
- Articulations of the foot.
- Joints of hand.
- Wrist Joints.
- Axillary articulations.
- Temporomandibular joints.
- Knee Joints.

3.2.2 The joint space

The distance between the involved bones of the joint is called the joint space. The normal joint space is at least 2 mm in the hip, at least 3 mm in the knee, and 4-5 mm in the shoulder joint. So the normal joint space is between 1.5 and 4 mm (Drake et al., 2009).

3.2.3 Factors of joint stability

A number of factors interact to confer stability while permitting motion in active human joints.

- The shape of the component parts. For example, in the hip joints, the weight-bearing drives the femoral head into a relatively deep socket, the acetabulum (Frederick A. Matsen III, 2011).
- Ligaments guide and align normal joints through their range of motion. For example, in the knee joints, these strong relatively inelastic ligaments limit articular motion of flexion and extension (Frederick A. Matsen III, 2011).
- Muscles and tendons provide the axes of motion and the flexibility that is required. For example, in the shoulder joints, the rotator cuff muscles stabilize the articular surfaces of the shoulder for effective shoulder motion (Frederick A. Matsen III, 2011).

3.2.4 The types of joints with the example of architectural projects

3.2.4.1 Fibrous joints

Immovable joints with thick connective tissue categorized into three types:

- **Sutures joints:** sutures are nonmoving joints like bones of the skull (Picture 3-10) that have serrated edges that are locked together with fibers of connective tissue.

Pic 3-10: Sutures joints in skull

- Example of sutures joint

As in the human body, buildings have different joints. Consciously or unconsciously, we see a different type of joints in buildings that they are partly the same as human body Sutures joints. These are some examples of joints that partly the same as sutures joint (Picture 3-11).

Pic 3-11: Examples of sutures joint in buildings

- **Gomphoses joints:** the immovable joints like the joints between the teeth and the mandible or maxilla (Picture 3-12).

Pic 3-12: Gomphoses joint in teeth

- Example of Gomphoses joint

Pic 3-13: Examples of Gomphoses joint in buildings

These are some examples of joints that we use in our buildings that partly the same as Gomphoses joint (Picture 3-13).

- **Syndesmosis joints:** a little movements joints such as the distal joint between the tibia and fibula is an example of a syndesmosis joint that is connected with ligaments to each other (Picture 3-14a) (“Bones Come Together: Types of Joints in the Human Body,” n.d.).

- Example of Syndesmosis joint

No example has been found yet

3.2.4.2 Cartilaginous joints

This kind of joints has two types.

- **Synchronous joints:** an immovable and temporary joint that the cartilage is converted into bone before adult life. For example, the joint between the first pair of ribs and the sternum (Picture 3-14b) and the epiphyseal plates in long bones (Picture 3-14c) (“Types of Joints,” n.d.).

- Example of synchronous joint

No example has been found yet

- **Symphysis joints:** a fibrocartilaginous fusion between two bones that are slightly movable, permanent and growing together of parts or structures. Such as the hip bones (Picture 3-14d) and the vertebrae connected with intervertebral discs (Picture 3-14e) (“Bones Come Together: Types of Joints in the Human Body,” n.d.).

Pic 3-14: Cartilaginous joints (a) Syndesmosis joints between the tibia and fibula, (b) Synchondrosis joint between the first pair of ribs and the sternum (c) Epiphyseal plate (d) Symphysis joints in hip bones (e) Symphysis between vertebrae

- Example of symphysis joint

For this type of human body joint, we have some buildings that are inspired by the human body. Some of them are similar in shape and some are shaped according to its structure.

Eternal Tower (China), Mount Tai has been prominent of cultural and religious significance in the state of Shandong China. It is one of China's five sacred mountains with concepts of sunrise, birth, and rebirth. The bridge is composed of three large rocks and several small rocks on the top of a valley. The person believes that it exists from the last ice age (Picture 3-15).

Pic 3-15: Eternal Tower (China)

Turning torso (Sweden), Turning Torso is the first twisting tower based on a sculpture of the human body by Calatrava (Picture 3-16) (Figure 3-1) ("Types of Joints," n.d.). This is a solid

Pic 3-16: Turning torso (Sweden)

Fig 3-1: Turning torso based on human body

immobile with twisted 90 degrees clockwise around the vertical core building is constructed in nine segments of five-story floors and supported by an exterior steel framework.

The same as the other joints; these are some examples of joints that partly the same as Symphysis joint such as buildings earthquake stabilization (Picture 3-17) and stabilization construction equipment (Picture 3-18).

Pic 3-17: buildings earthquake Stabilization

Pic 3-18: Stabilization construction equipment

Examples of behavior that can be spine-like are the connection between the wagons of trains or buses (Picture 3-19).

Pic 3-19: The joints of Bus and Train cabins

3.2.4.3 Synovial joints

This type of bone joints contains six types, which are described below.

- **Ball and Socket joints:** the largest joints in the body that we have in the shoulder and hip joints (“Types of Joints,” 2012) is shaped as follow:

The end of the upper arm and leg bone is rounded, looks like a half ball. It fits into a cup-like socket. This type of joints provides the greatest range of 360-degree movement such as moving backward, forward, and to the side and rotation of the long bones. This rotation is providing a remarkable range of complex movement for the arms and legs (Picture 3-20) (John, 2017).

Pic 3-20: Ball-and-Socket Joints in hip joints

- Example of ball and socket joint

Hub joints, the ball and socket joints simplify construction and enable adjustment of the basic structure (Picture 3-21).

Ball Joints, this is a variant of the simple ball and socket joint where with gaps in the walls of the socket to allow to rotate more than 90-degree angle in four points (Picture 3-22).

Pic 3-21: Hub joints

Pic 3-22: Ball Joints

The same as the other joints, these are some examples of joints that partly the same as ball and socket joint such as building a connection (Picture 3-23a), nuzzle joints (Picture 3-23b), artificial joint of the human body (Picture 3-23c) and mechanical connection (Picture 3-23d).

(c)
Pic 3-23: (a) Building connections, (b) Nuzzle joints, (c) Artificial joint of human body, (d) Mechanical connections

- **Hinge joints:** this joint acts are similar to hinges on doors. They roll just back and forth movement not a side-to-side or lateral movement. Their differences from hinges on doors is the role of ligaments to connect two moving bones. This type of joints include the elbows, the knees, the middle and end joints of the fingers (Picture 3-24) (John, 2017).

Pic 3-24: Hinge Joints example

- The example of Hinge joint

Hinge joints don't bring as much movement as ball and socket joints, but they're much simpler to make.

The structure based on the mobility of the hinge joint (Picture 3-25).

Pic 3-25: The structure based on the mobility of hinge joint

The structure based on the stability of the hinge joint (Picture 3-26).

Pic 3-26: The structure based on the stability of hinge joint

Making the building based on hinge joint (Picture 3- 27).

Pic 3-27: Making the building based on hinge joint

Making the bridge based on the hinge joint:

Gates head Millennium Bridge, UK, it contains a deck for pedestrians and cyclists and a supporting arch. Spanning the River Tyne, this bridge is nicknamed because of its eye-shaped in motion. The bridge rotates as a single structure, with the arch lowering and the deck rising, counterbalancing each other to form a pathway for water traffic (Picture 3-28).

Pic 3-28: Gates head Millennium Bridge, UK

Madison Street Bridge, a bascule bridge over the Chicago River in Chicago, IL, today, there are a number of bridges in Chicago over its river, and all of them are moveable with attractive structures (Picture 3-29).

Pic 3-29: Madison Street Bridge, a bascule bridge over the Chicago River in Chicago, IL

Hörn Bridge, Germany, it is a folding “N” shape bridge over Germany’s Horn River. It is Bascule Bridge but it initially had many malfunctions (Picture 3-30).

Pic 3-30: Hörn Bridge, Germany

Foryd Harbor Bridge, UK, this modern pedestrian bridge is in the coast of Wales. It operates a bit like a sailing boat with a mast, cables, rigging system and pulley mechanism (Picture 3-31).

Pic 3-31: Foryd Harbor Bridge, UK

Slauerhoffbrug, Netherlands, this tail bridge with the nickname of the Flying Drawbridge is yellow and blue automatic bascule bridge that can quickly raise the deck to pass water traffic (Picture 3-32).

Pic 3-32: Slauerhoffbrug, Netherlands

- **Pivot joints:** pivot joints (rotary joints) are facilitated by the rotation of one bone around another. This joint is formed by the meeting of two bones, a bony cylinder and a ring-like structure constructed of bone and ligament such as the base of the skull (cause rotating of the

head by joining the first vertebra of the spine to the second vertebra) and radius bones in the forearm (Picture 3-33) (“Types of Joints,” 2012).

Pic 3-33: Pivot Joints in radius bones in the forearm

- The example of Pivot joints

Puente de la Mujer, Argentina, this rotating pedestrian bridge in Puerto Madero, Buenos Aires, has fixed two sections and a 90° rotating middle section in order to pass water traffic (Picture 3-34).

Pic 3-34: Puente de la Mujer, Argentina

River Hull Footbridge, UK, this pedestrian bridge in Hull, Yorkshire, is also a movable bridge. It can be rotated around 90° while pedestrians are still on it. Moreover, to create fascinating light show it has turning on and off lights during its rotating at night (Picture 3-35).

Pic 3-35: River Hull Footbridge, UK

El Ferdan Railway Bridge, Egypt, it is Egypt Railway Bridge, over the Suez Canal. This is the world's longest swing bridge with 335m. It takes a long time, 30 minutes, to open or close. It closed when the train needs to cross the canal (Picture 3-36).

Pic 3-36: El Ferdan Railway Bridge, Egypt

- **Ellipsoidal joints:** ellipsoidal joints (condyloid joints) permit angular and bending movements with limited rotation. These joints are more similar to ball-and-socket joints (Scheve, n.d.).

This node isn't accessible because of its position between the oval projection, the condyle of bone and hollow space available. This fits together in such a way to allow extension, flexion, movement, and lateral bending but it cannot rotate freely ("Types of Joints," 2012). Joints in the wrist, hands, and feet and index finger are the examples of ellipsoidal joints (Picture 3-37) (John, 2017).

Pic 3-37: Ellipsoidal Joints

- The example of Ellipsoidal joints

A bascule bridge, it is an upward swing bridge for providing a way for boat traffic. It is the French name for showing a balance scale. Bascule bridges open quickly and require little energy to operate so they are the most common type of movable span (Picture 3-38).

Pic 3-38: bascule bridges

Ashtabula Lift Bridge, this is lift bridge (the West Fifth Street Bridge) in a crucial location in the harbor of Ashtabula, Ohio (Picture 3-39).

Pic 3-39: Ashtabula Lift Bridge bascule bridges

- **Gliding joints:** gliding joints (plane joints) are combined with two bone plates that glide against one another such as the vertebrae of the spine and the eight small bones of the wrist and ankles (Picture 3-40).

A gliding joint makes a wide range of motion and holds joints steady while they point upward and then waving side-to-side. (Scheve, n.d.; “Types of Joints,” 2012).

Pic 3-40: Gliding joints between the eight small bones

- **The example of gliding joints**

There are some different model of gliding joint in building junction (Picture 3-41).

Pic 3-41: The different model of gliding joint in building junctions

- **Saddle joints:** the joint at the base of the thumb is the only saddle joint in the human body (Picture 3-42). The saddle joint allows the thumb to move toward the palm and grasp objects between your thumb and fingers (John, 2017).

Saddle joints are formed when the two concave and convex surfaces bones meet each other. The saddle joint must allow for movement in sliding flexion, extension, abduction, adduction, and some rotation ("Types of Joints," 2012).

Pic 3-42: saddle joint in thumb

- The example of gliding joints

The wooden houses are made based on the saddle joints (Picture 3-43)

Pic 3-43: The wooden houses

The same as the other joins, these are some examples of joints that partly the same as Saddle joints such as artificial joint (Picture 3-44) and bicycle saddle (Picture 3-45).

Pic 3-45: Artificial joint

Pic 3-44: Bicycle saddle

3.2.5 Joint actions

Joints provide different action by muscles. There are six types of joint action (“How do the musculoskeletal and cardiorespiratory systems of the body influence and respond to movement?,” n.d.) (Table 3-3):

Table 3-3: Joint actions

Joint action	Movement description	Example of movement
Flexion	decreasing joint angle & bending the limbs at a joint	Biceps Curl
Extension	increasing joint angle & straightening limbs at a joint	Triceps Extension
Abduction	movement away from body centerline	Lateral Raises (Deltoids)
Adduction	the movement toward body centerline	Horizontal Flyes (Pectorals)
Rotation	rotation about an axis & a circular movement around a fixed point	Twisting the Arm
Circumduction	360 degree rotation	Arm circle around

3.3 Human anatomy flexibility

Flexibility is the ability to stretch a joint according to the limit of its range of movement (Figure 3-2) (Medlej, 2014).

Fig 3-2: Human Anatomy Flexibility

3.3.1 General flexibility facts

There are some general flexibility facts based on their action:

- The female bodies have a tendency to more flexibility and the male body has more muscle power.
- Having the flexibility of one joint does not mean having flexibility in others.
- Having more muscles means having less flexibility. Bodybuilders have the most limited range of movement for all the joints because their muscles don't stretch besides having bulging muscles.
- Having more flexibility means having less muscle, so those muscles are often very fragile because they are so thin (Medlej, 2014).

3.3.2 Types of flexibility

In the human body, we have faced three different types of flexibility: passive, active, and dynamic.

- Passive flexibility is how much you can stretch your body with the outer reason such as the weight of your body or by a partner and without using muscle in this case (Figure 3-3).

Fig 3-3: Active, Passive and Dynamic Flexibility

- Active flexibility is how much you can stretch and freeze your parts of the body with your muscles.
- Dynamic flexibility is how much you can stretch your parts of the body by force or swinging it for a short time. This is usually happened during fights or actions because of their speed. It sometimes makes tearing muscles (Medlej, 2014).

3.3.3 The range of movement for some joint

We are going to discuss some joints as follow based on their range of motions (Figure 3-4):

Fig 3-4: Range of Movements

- The Neck

The neck consists of last seven vertebrae in the spine. They are separated but stacked each other with something like a cushion, so their movements become limited.

Fig 3-5: The Neck Flexibility

Flexion: the chin can touch the sternum with the range of 70-90 degrees.

Extension: the neck and jawlines make a single curve around 55 degrees.

Lateral bending: the neck could bend to 35 degrees but the ear cannot touch the shoulder.

Rotation: the neck on its own turns 70 degrees for each direction. To look further back, we have to engage the upper body too (Figure 3-5) (Medlej, 2014).

- The upper body (Thoracic spine)

The thoracic spine is included in most flexibility training. This part of the spine has very limited movement.

At rest: the spine has a normal shape of S-shape.

Flexion: this part could have flexion up to the vertebra at the base of neck juts out.

Extension: the extension of this part continues up to the space between the shoulder blades sinks. In the correct version, when you want to bend from the back, you can see that a number of joints engage to make a whole arch back.

Rotation: rotating the upper body by keeping the hips fixed is 30-40 degree in each direction (Figure 3-6, 3-7) (Medlej, 2014).

Fig 3-7: The Upper Body (Thoracic Spine) Flexibility

Fig 3-6: The Upper Body (Thoracic Spine) Flexibility

- The lumbar spine (Lower back)

It is made of the five vertebrae that emerge from the pelvic bone, so its range of movement is quite limited, but it helps us to have considerable motion in other parts (Figure 3-8) (Medlej, 2014).

Fig 3-8: The Lumbar Spine (Lower Back)

- The ankle

The ankle joint the same as the wrist has a limited range of rotation (Figure 3-9) (Medlej, 2014).

Fig 3-9: The Ankle Flexibility

- The shoulder joint

Of all the joints in our body, this joint has the biggest range of movement and rotate 360° continuously because of its socket articulation. It also can pop out of place easily if pushed too hard (Figure 3-10) (Medlej, 2014).

Fig 3-10: The Shoulder Joint Flexibility

- The elbow

The elbow works as a door hinge; it opens in one direction and meets a stop.

Extension: it could extend up to 180 degrees but some people's elbows have hyperextension.

Flexion: this joint can close all the way, but its angle depends on the size of muscle mass (Figure 3-11).

No internal rotation: these are some additional movements that don't belong to the elbow proper, but can be treated here. These movements happen because of rotation of the wrist (Medlej, 2014).

Fig 3-11: The Elbow Flexibility

- The hip joint

It behaves the same as the shoulder joint because of its socket articulation, but more limited.

Fig 3-12: The Hip Joint Flexibility

Flexion: this range is valid when the knee is bending or kicking and the angle of this flexion depends on the muscle mass.

Extension: turning the foot little outward causes the leg to swing back a few more degrees (around 30 degrees).

Abduction: because of the position of the joint, turning the toes outward brings a considerably bigger range than forwarding (Figure 3-12) (Medlej, 2014).

- The knee

The knee has the same role as the elbow:

Flexion: the same as the elbow, we have a flexion but its angle depends on legs muscle mass.

Extension: the knee could extend to 195 degrees but it doesn't hyperextend.

Internal rotation: it could rotate about 10 degrees and this becomes a greater degree of the rotation of the hip joint (Figure 3-13) (Medlej, 2014).

Fig 3-13: The Knee Flexibility

- The wrist and fingers

The wrist's range of motion is normally front and back with so little rotate to the sides.

Fig 3-14: The Wrist and fingers flexibility

Fingers are quite flexible in extension. They can only extend a little unaided and it requires much tension to bend back up to 90°. They can also bend back individually when the rest of the hand is relaxed (Figure 3-14) (Medlej, 2014).

3.4 Preliminary conclusions

Nowadays, Bionics structures have brought about a revolution in the development of technology and it is still being developed (Taghizadeh & Bastanfard, 2012) by the combination of realities based on biology, mathematics, construction and mechanics to make algorithmic geometry, fractals and chaos theory, intelligent materials and systems, opening and closing mechanisms (Salsabili et al., 2016), genetic algorithms, , algebra fractal, and neural networks. The bionic architecture promote sustainable development of complex systems by increasing the efficiency of resources and adapting to sustained life(El Ahmar, 2011).

There are extensive examples in nature for modeling and many of them have already been used. In the meantime, but taking the advantages of the human body especially in the field of construction has been neglected. At first glance, it might be thought that the human body is very complex, but having studied this point, we can conclude that in spite of our imagination, it is much understandable and usable structure. In addition, there have been many studies about its' behavior in both aspects of anatomical and mechanical behavior.

The human body contains three main parts of the skeletal system (it contains human joints as the framework of the human body), muscles (it makes the human body moveable) and brain system (it changes the human body to the conscious system). According to this study, the human joints are the good examples of modeling because of both opposite aspects of stability and mobility, different Joint shapes, aesthetic form, different Joint features, comfort form, and light structures that they have. On the other hand, they are more different with respect to each other in their roles of moving as their functions. So, they are good examples of modeling to make different structures with different capacity and functions. As described in this chapter, there are many examples of these types of joints in building and making structures conscious or unconsciously. Furthermore, they can be used in the construction of structures by exploiting their functional capacities. However, they can be used more for further structure by having their capacities.

CHAPTER IV: THE ANATOMICAL ROLES OF THE HUMAN SPINE

4.1 Anatomy of the human spine

Spine, vertebral column or backbone is one of the most important parts of our axial skeleton. There is a vertebral column in about 50,000 species of animals, but the human vertebral column is the most-studied.

The spine bones (33 individual bones) are stacked one on top of the other (Marco A. W. Rex, 2002). They make our body structure, body-supporting, body-standing, and body-up righting, body movement and body-twisting freely, body-bending with flexibility and protecting our spinal cord ("Anatomy and Function," n.d.). The spinal column has four main sections-the cervical spines (The first seven vertebrae), the thoracic spine (The mid-back consist of 12 vertebrae), the lumbar spine (usually five vertebrae), and sacrum (five vertebrae) with coccyx (four vertebrae) (Picture 4-1) ("Anatomy and Function," n.d.). The vertebrae of the cervical, thoracic and lumbar spines are independent bones, and generally quite similar (except two beginner vertebrae of the spine (atlas and axis)), but the vertebrae of the sacrum and coccyx are unable to move independently.

Each part of the spine has some differences in the form, function, and attachment to adjacent structures (Picture 4-2) (Thomas Scioscia, 2017). This column supports about half of the weight of the body itself and the other half is supported by the muscles (Thomas Scioscia, 2017).

Pic 4-2: The vertebrae of spine Pic 4-1: The shapes of vertebra from top and side

4.2 Spine functionality

4.2.1 The functions of the spinal column

The whole functions of the spinal column consist of:

- Supporting the major weight of the body.
- Supporting the head, trunk, and UEs against the gravity.
- Protecting the spinal cord.
- Absorbing shock.
- Providing a stable structure to have an upright posture (Picture 4-3) (“Dutton_Chapter_23_Cervical_Spine,” 2017).

Pic 4-3: Spine Function

4.2.2 The functions of each side of the spine

The functions of each side of the spine include:

- Anterior Pillar: it includes vertebral bodies and intervertebral disks, so they provide hydraulics, weight-bearing ability, and shock-absorption.
- Posterior Pillar: it includes articular processes, facet joints, transverse processes, and spinous processes. They allow spinal movement and bring the attachment for posterior musculature (“Dutton_Chapter_23_Cervical_Spine,” 2017).

4.2.3 The motion of the spine

According to White and Panjabi, there is the flexibility of motion in six degrees of freedom:

- Flexion and extension.
- Left and right side bending.
- Left and right rotation (Picture 4-4) (Pankaj N Surange, n.d.)

Pic 4-4: The motions of spine

4.3 Spinal curves

From the side, an adult spine is a natural S-shaped curve with three curves:

- Cervical curve: the concave curve of the neck.
- Thoracic curve: the convex curve of sacral regions.
- Lumbar curve: the concave curve of low back regions (Picture 4-5) (“Anatomy and Function,” n.d.).

The thoracic curves are primary curves as they are present in the fetus ages. The cervical and lumbar curves are secondary and are developed after birth. The cervical curve forms when the infant is able to hold up its head (at three or four months) and to sit upright (at nine months). The lumbar curve forms later from twelve to eighteen months, when the child begins to walk. (Picture 4-6).

Pic 4-5: The curves of spine Pic 4-6: The curves of spine during the human growing up

4.3.1 Cervical spine (Neck)

The cervical spine is the first seven vertebrae in the spine that get bigger from the base of the skull (Thomas Scioscia, 2017). This lordosis curve (concave curve) is much more mobile than both of the other spinal regions (“Anatomy and Function,” n.d.).

The atlas and the axis, the first two vertebrae in the cervical spine, differ from the other vertebrae because they are designed for rotation (“Anatomy and Function,” n.d.).

- The atlas (the first cervical vertebra) does not have a vertebral body, but it has a thick forward arch and a thin back arch (“Anatomy and Function,” n.d.).
- The axis has a bony knob (the odontoid process) sticks up through the hole in the atlas. There is a special ligament between the atlas and the axis to allow a great deal of rotation (“Anatomy and Function,” n.d.).

The cervical spine is more flexible, but much at risk for injury due to the limited muscle for supporting the cervical area and supporting the weight of the head (an average of 15 pounds) (“Anatomy and Function,” n.d.). The most rotation of the cervical spine belongs to the top two vertebrae and the most flexion-extension movement belongs to C5-C6 and C6-C7 (Picture 4-7) (Thomas Scioscia, 2017). Against the rest of the spine, there are special openings in each vertebra in the cervical spine for blood vessels (arteries) to go to the brain (“Anatomy and Function,” n.d.).

Pic 4-7: The cervical vertebrae

4.3.2 Thoracic spine (Mid Back)

The middle 12 vertebrae make the thoracic curve. The ribs are connected to this curve and form the back wall of the thorax. The thoracic spine's curve is convex (kyphosis) (Picture 4-8). This part has little movements because of thin intervertebral discs and attaching to Ribs, but it is too strong to protect the vital organs and it has rare injuries (“Anatomy and Function,” n.d.).

Pic 4-8: The thoracic vertebrae

4.3.3 Lumbar spine (Low Back)

The five lowest vertebrae are the lumbar spine. The lumbar spine has a concave (lordosis) curve. The vertebrae and the spinal canal in this area are the largest of the whole spine (Picture 4-9). Most of your weight bearing and body movement takes place in this part so the back pain is a very common (“Anatomy and Function,” n.d.).

Pic 4-9: The lumbar vertebrae

4.3.4 The sacral region (Bottom of the spine)

This part consist of two parts:

- Sacrum: this triangle shape bone (consist of 5 vertebrae) is the connection between the spine and pelvis (Thomas Scioscia, 2017).
- Coccyx (the tailbone): it is at the bottom of the sacral region (Picture 4-10) (Thomas Scioscia, 2017).

Pic 4-10: The sacral and coccyx vertebrae

4.4 Structures of the human spine

A spinal segment is consist of two vertebrae that are attached by ligaments together (Picture 4-11a), with a soft disc between them. The facet joints fit the two vertebrae to allow them for movement, the ligaments to join the vertebrae to each other, the muscles to make the movement of spine, and the neural opening between the vertebrae make space for the nerve roots to travel freely from the spinal cord to the body (“Anatomy and Function,” n.d.).

The vertebra consists of these items (Picture 4-11b):

- Vertebrae.
- Intervertebral Discs.
- Facet Joints.

- Ligaments.
- Muscles.
- Spinal cord.
- Spinal nerves (“Anatomy and Function,” n.d.).

Pic 4-11: (a) The vertebra structures from top, (b) The vertebra joint from side

4.4.1 Vertebrae

According to the previous parts, the spine consists of 33 individual bones with different shapes and characters that interlock with each other (Picture 4-12). The vertebrae of the sacrum and coccyx are fused but the top 24 bones are moveable (Marco A. W. Rex, 2002).

- Cervical (neck) are numbered C1 to C7 for supporting the weight of the head and the most range of its motions (due to the first two vertebrae (atlas and axis)) (Marco A. W. Rex, 2002).
- Thoracic (mid back) are numbered T1 to T12 for holding the rib cage and protecting the heart and lungs. It has a limited range of motion (Marco A. W. Rex, 2002).
- Lumbar (low back) are numbered L1 to L5 for bearing the weight of the body. These vertebrae are too large in size for absorbing the stress of lifting and carrying heavy objects (Marco A. W. Rex, 2002).
- Sacrum with five fused vertebra works for connecting the spine to the hip bones (iliac) and making the pelvic girdle (Marco A. W. Rex, 2002).
- Coccyx region (tailbone) with four fused bones provides attachment of ligaments and muscles to pelvic (Marco A. W. Rex, 2002).

Pic 4-12: the spine vertebrae

4.4.1.1 Vertebra parts

Besides every regional feature, each vertebra has these three functional parts:

- A drum-shaped body, for bearing weight and enduring compression (purple part).

Pic 4-13: (a) The vertebrae parts from top, (b) the vertebra parts from side

- An arch-shaped bone, for protecting the spinal cord (green part).

- Star-shaped, for attaching to muscles and ligaments (tan). This part has seven activities such as the spinous process, two transverse processes, two superior facets, and two inferior facets (Picture 4-13a, 4-13b) (Marco A. W. Rex, 2002).

4.4.1.1.1 Similar examples of the vertebra

What was discussed about the structure of a vertebra shows that the vertebrae of the spine are similar to the Lego pieces (Picture 4-14) which are fastened together with modular and similar shapes (Salsabili, 2014)?

Pic 4-14: The Lego pieces

4.4.1.1.2 A diagram in place of the vertebra

If we want to show the vertebra with Lego pieces roles, we received these types of diagrams (Figure 4-1) (Salsabili, 2014).

Fig 4-1: The diagram of imagination of vertebrae as Lego pieces

4.4.1.2 Vertebra component

Vertebrae, like all bones, have two different parts.

- The outer shell (cortical bone) is hard, strong and having much dense (contributing about 80% of the weight of a human skeleton) (Buffrenil & Mazin, 1990).
- The inside (cancellous bone, trabecular or spongy bone) is soft, spongy, less dense and more flexible with a higher surface area ("Anatomy and Function," n.d.). So its function is being aligned towards the mechanical load distribution (Gdyczynski, Manbachi, Hashemi, Lashkari, & Cobbold, 2014) (Picture 4-15a).

The vertebrae are stacked one on top of the other to make the hollow tube with their rings for the spinal cord to pass through. Besides, each vertebra is connected to the others by groups of ligaments and muscle is connected to the vertebra by tendons ("Anatomy and Function," n.d.).

According to the results of the studies by Wu and Chen, pressure on the vertebral body is always more in the front part (anterior) than the back part (posterior).

Pic 4-15: (a) The component of vertebrae, (b) The trabecular systems of vertebra

The vertical, horizontal and inclined trabecular systems (Picture 4-15b) which are consistent with the pressures on the vertebral body are found in spongy bone. Vertical systems contribute to body weight bearing and resist against compressive forces. Other trabecular systems are resistant to cutting forces (Salsabili, 2014).

4.4.1.2.1 Similar examples of the vertebra

We see a typical architectural structure in the human spine because of loading the variety of bars with its vertebra (Smit, Odgaard, & Schneider, 1997).

The important point is that trabecular system with a cortical part similar to the structural systems transmit vertical forces to the next vertebra by itself and works against cutting forces as well. Therefore, it is possible to design a vertebra as follows with different kinds of present structural systems (Picture 4-16) (Salsabili, 2014).

Pic 4-16: The structure of building

4.4.1.2.2 A diagram in place of the vertebra

So, we could simulate the vertebra with each type of building structures (Figure 4-2) (Salsabili, 2014).

Fig 4-2: The diagram of imagining of vertebrae as structures of buildings

4.4.2 Intervertebral discs

The vertebral disc in the spine act as a shock absorber between adjacent vertebrae like a cushion or a radial car tire. Its role is holding the vertebrae together, allowing for slight mobility (Thomas Scioscia, 2017), and keeping the bones from rubbing each other.

- The outer ring (annulus) contains crisscrossing sheets of collagen fibrous bands. The annulus is too strong with strong ligaments to connect vertebra and help the disc's center to keep intact (Marco A. W. Rex, 2002), and it helps the vertebral bodies to have the elastic resistance of the gel-filled nucleus.
- The Inside of the disc is a loose network of fibers in the gel-filled center (nucleus) (Marco A. W. Rex, 2002). It is made up of tissue and water so it is moist and mushy to be as the main shock absorber the same as the rolls of a waterbed mattress ("Anatomy and Function," n.d.). The gel-filled nucleus consists mostly of fluid. This fluid is absorbed during the night when you lie down and is pushed out during the day when you move upright (Picture 4-17a) (Marco A. W. Rex, 2002).

Intervertebral disk is made up 20-30% of the height of the column of the spine and its thickness are different (3mm in the cervical region, 5mm in the thoracic region to 9 mm in the lumbar region) ("Spine 08_Revised," n.d.). The spinal disc is composed of fluid with few nerve endings and no blood supply. So, the disc can't repair itself (Thomas Scioscia, 2017), and by aging our discs lose the ability to reabsorb the fluid and become brittle, flat and shorter (Marco A. W. Rex, 2002).

The intervertebral disc provides as a whole:

- Shock absorption.
- Movement of vertebrae: Flexion, Extension, Lateral Flexion (Picture 4-17b), Rotation and Shear.
- The separation between the vertebrae.

- To allow passage of nerve roots through the intervertebral foramina (“Dutton_Chapter_23_Cervical_Spine,” 2017).

Pic 4-17: (a) Intervertebral disk from top and side, (b) The flexion and extension of vertebrae

The compositions of nucleus and annulus fibers are similar, in the sense that they both have water, collagen, and proteoglycans. However, the relative proportion of these materials and types of collagen in the two layers are different. Water and proteoglycans density is the highest in the nucleus and the lowest in the external fibers. On the contrary, the density of collagen fibers in the external annulus fibers is the highest and in the nucleus is the lowest. The system and material properties of the discs lead to the viability of the discs (which have the power of self-reconstruction in case of injury) and their flexibility (Salsabili, 2014).

The annulus collagen fibers are made of concentric of 10-12 layers sheets (lamellae) (Picture 4-18) with 120 (or 45-65) degrees and opposite directions (“Spine 08_Revised,” n.d.).

Pic 4-18: The lamellae of Intervertebral disk

4.4.2.1 Similar examples of intervertebral discs

In general, what is observed in the behavior of inter-vertebral discs is similar to what is seen in the behavior of intercity trains, wagons, cushion, car tire, and double-decker buses (Picture 4-19) which have air mattresses that are responsible for transmitting pressure to adjacent carriages? Therefore, this system may be simulated by considering an air or silicon mattress, which has elasticity and can transmit pressure from one module to another and can distribute it in a surface and sometimes if needed it can be repaired, injected or optimized. In order to distribute the load or pressure on the surface, a transition structure system must be used beneath and above the mattress (Salsabili, 2014).

Pic 4-19: The behavior of intercity cushion, car tire and double-decker buses are the same as intervertebral disc

4.4.2.2 A diagram place of intervertebral discs

Hence, for better load transfer, the structure must have better adaptability and correspondence with the disc in it so that it spreads the load more uniformly. On one hand, since the most pressure is transmitted from one vertebra to another by the nucleus, and on the other hand, the annulus fibers transmit pressure from the nucleus, a nucleus can be utilized (Figure 4-3) (Salsabili, 2014).

Fig 4-3: The diagram of imagining of Intervertebral disk with vertebrae

4.4.3 Facet Joints

The facet joints are synovial joints for linking the vertebrae together and giving them the flexibility to move against each other (“Anatomy and Function,” n.d.) and having back motion such as flexion, extension, rotation, and lateral bending (Thomas Scioscia, 2017). Each vertebra has four facet joints, one pair that connects to the vertebra above (superior facets) and one pair that connects to the vertebra below (inferior facets) (Picture 4-20) (Marco A. W. Rex, 2002). Facet joints are synovial joints so they have articular cartilage and a slick spongy material that allows the bones to glide against one another without much friction (“Anatomy and Function,” n.d.).

Pic 4-20: The facet joints

4.4.3.1 A diagram place of facet joints

This way, joints are similar to logo beads which cause modules not to rotate on each other (Figure 4-4) (Salsabili, 2014).

Fig 4-4: The diagram of facet joints in vertebrae

4.4.4 Ligaments

The ligaments are strong densely packed collagen fibrous bands with the limited blood supply that hold the vertebrae together, stabilize the spine, and protect the discs (Marco A. W. Rex, 2002). Ligaments connect bone to bone and tendons attach muscle to bone (Picture 4-21) (Stewart G. Eidelson, n.d.).

The six major ligaments of the spine are:

- ligament flavum (LF), this is a thick and elastic ligament that is located posterior to the spinal canal which is drawn from the second cervical vertebra to the sacrum and is

connected to the lamina blade of adjacent vertebrae. Although the maximum strain in this ligament is during flexion, it is under constant tension even in a neutral state (Intersegmental) (Salsabili, 2014).

- Anterior longitudinal ligament (ALL), throw along the vertebral bodies from the top to the bottom of the spinal column to prevent excessive movement of the vertebral bones of the spine (Stewart G. Eidelson, n.d.). It is compressed in flexion and is extended or stretched too. It has twice the power of the posterior longitudinal ligament (Intersegmental) (Salsabili, 2014).

- Posterior longitudinal ligament (PLL), throw along the vertebral bodies from the top to the bottom of the spinal column (on the opposite side of the anterior longitudinal ligament) to prevent excessive movement of the vertebral bones of the spine (Marco A. W. Rex, 2002). This ligament is stretched in flexion with the highest strain and is loosen in extension (Intersegmental) (Salsabili, 2014; Stewart G. Eidelson, n.d.).

These two types of ligaments (ALL and PLL) have two layers. The first layer is long surface fibers and joins multiple vertebrae and the other layer are short inter-vertebra fibers that only join two vertebrae (Salsabili, 2014).

Pic 4-21: The ligaments place

- Interspinous ligaments (ISL), these ligaments stick to the edge of the adjacent spinous processes and cover them. These ligaments are loose during extension and in anterior flexion resist by the distancing, the spinous process when bending forward occurs. So, it has less potential power against tension or strain than the anterior longitudinal, the posterior longitudinal ligaments, and ligamentum flavum (Salsabili, 2014).

- Supraspinous ligaments (SSL), this ligament connects the tips of spinous processes. This ligament is stretched in flexion and is sharply stretched in bending forward (Salsabili, 2014).

- Intertransverse ligaments (IL), this ligament is pulled from one side and is pressed from the other side during lateral flexion (Salsabili, 2014).

4.4.4.1 Similar examples of ligaments

Ligaments only work in stretches or tensions like cables (Picture 4-22a) (Salsabili, 2014).

In fact, ligaments are like the string of prayer Tasbih beads (Picture 4-22b) which connects the beads and is always under pressure (Salsabili, 2014).

(a)

(b)

Pic 4-22: (a) Cable, (b) Prayer Tasbih beads

4.4.4.2 A diagram in place of ligaments

So, the vertebrae are connected to each other by a series of ligaments to show an integral behavior (Figure 4-5) (Salsabili, 2014).

Fig 4-5: The diagram of ligaments in vertebra

4.4.5 Muscles

Muscles with their striped appearance are named based on their shape, function, combination, and location. They are further categorized according to their function such as extension, flexion, or lateral (Picture 4-23) (Tables 4-1, 4-2, 4-3, and 4-4) (Bridwell & Rodts, n.d.).

- 1- The extensor muscles (back) are attached to the posterior of the spine and enable standing, lifting objects and stabilizing your spine (Marco A. W. Rex, 2002). These muscles include the large paired muscles in the lower back.
- 2- The flexor muscles (front) are attached to the anterior of the spine (which includes the abdominal muscles) and enable flexing, bending forward, lifting, and arching the lower back. These muscles include the abdominal muscles (Thomas Scioscia, 2017).

3- The oblique muscles (both sides of the spine) are attached to the sides of the spine and help rotate the spine and maintain proper posture (Thomas Scioscia, 2017).

Pic 4-23: The most important muscles types in spine

Table 4-1: Types of vertebrae

Types of Vertebral Muscles	General Location
Forward flexors	Anterior
Lateral flexors	Lateral
Rotators	Lateral
Extensors	Posterior

There are different types of muscles based on their location:

Table 4-2: Cervical muscles

CERVICAL MUSCLES	FUNCTION	NERVE
Sternocleidomastoid	Extends & rotates the head, flexes vertebral column	C2, C3
Scalenus	Flexes & rotates the neck	Lower cervical
Spinalis Cervicis	Extends & rotates the head	Middle/lower cervical
Spinal Capitus	Extends & rotates the head	Middle/lower cervical
Semispinalis Cervicis	Extends & rotates vertebral column	Middle/lower cervical
Semispinalis Capitus	Rotates head & pulls backward	C1 – C5
Splenius Cervicis	Extends vertebral column	Middle/lower cervical
Longus Colli Cervicis	Flexes cervical vertebrae	C2 – C7
Longus Capitus	Flexes head	C1 – C3
Rectus Capitus Anterior	Flexes head	C2, C3

Rectus Capitus Lateralis	Bends head laterally	C2, C3
Iliocostalis Cervicis	Extends cervical vertebrae	Middle/lower cervical
Longissimus Cervicis	Extends cervical vertebrae	Middle/lower cervical
Longissimus Capitis	Rotates head & pulls backward	Middle/lower cervical
Rectus Capitus Posterior Major	Extends & rotates the head	Suboccipital
Rectus Capitus Posterior Minor	Extends head	Suboccipital
Obliquus Capitus Inferior	Rotates atlas	Suboccipital
Obliquus Capitus Superior	Extends & bends head laterally	Suboccipital

Table 4-3: Thoracic muscles

THORACIC MUSCLES	FUNCTION	NERVE
Longissimus Thoracis	Extension, lateral flexion of the vertebral column, rib rotation	Dorsal primary divisions of spinal nerves
Iliocostalis Thoracis	Extension, lateral flexion of the vertebral column, rib rotation	Dorsal primary divisions of spinal nerves
Spinalis Thoracis	Extends vertebral column	Dorsal primary divisions of spinal nerves
Semispinalis Thoracis	Extends & rotates vertebral column	Dorsal primary divisions of spinal nerves
Rotatores Thoracic	Extends & rotates vertebral column	Dorsal primary divisions of spinal nerves

Table 4-4: The lumbar muscles

LUMBAR MUSCLES	FUNCTION	NERVE
Psoas Major	Flexes thigh at hip joint & vertebral column	L2, L3, sometimes L1 or L4
Intertransversarii Lateralis	Lateral flexion of vertebral column	Ventral primary division of spinal nerves
Quadratus Lumborum	Lateral flexion of vertebral column	T12, L1
Interspinales	Extends vertebral column	Dorsal primary divisions of spinal nerves
Intertransversarii Mediales	Lateral flexion of vertebral column	Dorsal primary divisions of spinal nerves

Multifidus	Extends & rotates vertebral column	Dorsal primary divisions of spinal nerves
Longissimus Lumborum	Extends & rotates vertebral column	Dorsal primary divisions of spinal nerves
Iliocostalis Lumborum	Extension, lateral flexion of vertebral column, rib rotation	Dorsal primary divisions of spinal nerves

The movement of the spine is shaped by Joints (flexibility) and more than 30 Muscles and tendons (mobility) (Picture 4-24) (“Anatomy and Function,” n.d.; Stewart G. Eidelson, n.d.). Besides the muscles, ligaments work to support the spine, hold it upright, and control movement during rest and activity. Muscle is the only type of body tissue with the ability to contract. It becomes shorter and thicker during contraction. Some muscles work in pairs or as the opposite (Stewart G. Eidelson, n.d.). There are several layers of fibrous connective tissue (Fascia) for covering muscles. It extends beyond the muscle to the tendon that attaches the muscle to bone (Stewart G. Eidelson, n.d.). Muscles for working need energy (adenosine triphosphate (ATP)). ATP is produced as the burn of glucose (sugar) by mitochondria and blood vessels deliver the oxygen and nutrients for mitochondria to work (Bridwell & Rodts, n.d.).

Pic 4-24: The places of each muscles in spine

4.4.5.1 Similar examples of muscle

Considering what was mentioned about muscles, muscles function as an engine or mechanical lever (Picture 4-25). Which alters the shape and state of vertebrae. By placing each of them,

it is possible to make a structure movable in all directions. This depends on whether we want to have a moveable and dynamic or fixed stable structure (Salsabili, 2014).

Pic 4-25: the mechanical lever of window

4.4.6 Spinal cord

The spinal cord is a column of nerves for connecting our brains to the rest of our bodies, sending signals to other parts of the body and organs, and allowing controlling our movements.

The spinal cord consists of grey and white matter and a central cavity (central canal) (“Anatomy and Function,” n.d.).

Pic 4-26: The spinal cord and nerves

The spinal cord is about 18 inches long and the thickness of our thumb from the brainstem to the 1st lumbar vertebra within the protective spinal canal then it continues down through the spinal canal to our tailbone before branching off to your legs and feet between each of our vertebrae. The spinal cord serves as an information super-highway to relay messages between the brain and the body (Marco A. W. Rex, 2002).

The spinal cord branches off into 31 pairs of nerve roots, which exit the spine through small openings on each side of the vertebra called neural foramina (Picture 4-26). The two nerve roots in each pair go in opposite directions when traveling through the foramina ("Anatomy and Function," n.d.). The spinal cord comes off the base of the brain, runs throughout the cervical and thoracic spine, and ends at the lower part of the thoracic spine. Therefore, the spinal cord does not run through the lumbar spine (lower back). After the lower thoracic spine, the nerve roots exit the spine like a "horse's tail" (Thomas Scioscia, 2017).

4.4.7 Spinal nerves

The spinal cord is divided into spinal nerves (Thomas Scioscia, 2017). The spinal nerves act the same as "telephone lines," to carry messages back and forth between our body and spinal cord for controlling movement and sensation. These pairs of the nerve roots run through the vertebrae canals. According to our vertebrae, there are thirty-one pairs of spinal nerves branches. Each spinal nerve has two roots. The ventral (front) root carries motor messages from the brain and the dorsal (back) root carries sensory messages to the brain (Picture 4-27a).

Each branch of the spinal nerve has two motor and sensory fibers. The posterior primary antenna (the smaller one) for supplying the skin and muscles of the back of the body. The anterior primary antenna (the larger one) for supplying the skin and muscles of the front of the body (Picture 4-27b) (Marco A. W. Rex, 2002).

Pic 4-27: (a) The ventral and dorsal roots of spinal nerves, (b) Posterior and anterior antenna of spinal nerves

This pattern shows which spinal nerves are responsible for sensory and motor control of specific areas of the body.

- The nerves of the cervical spine go to the upper chest and arms.
- The nerves of the thoracic spine go to the chest and abdomen.
- The nerves of the lumbar spine reach the legs, pelvis, bowel, and bladder (“Anatomy and Function,” n.d.).

The spinal cord the same as the brain has three membranes (meninges) (Picture 4-28a).

- The inner membrane is the pia mater, which is exactly attached to the cord.
- The next membrane is the arachnoid mater.
- The outer membrane is the hard Dura mater.
- Between these membranes are spaces, and just the space between the pia and arachnoid mater is contained in cerebrospinal fluid (CSF).

4.4.7.1 Similar examples of spinal cord & nerve

Based on the spinal cord and nerves, we found that they have the rolls of the telephone line (Picture 4-28b) to send the data from their brain to its devices.

Pic 4-28: (a) The membranes of spinal cord, (b) The telephone line

4.4.8 The structure based on the anatomical roles of the human spine

As the results of this part, we found this structure based on the human spine without using muscles and spinal cords and nerves (Figure 4-6) (Picture 4-29). These findings suggest that the two motion segments of human lumbar spine can be used for making stable structures.

Fig 4-6: Final diagram of compacting the vertebrae, intervertebral disk, and ligaments together

Pic 4-29: Final modeling of spine structures without muscles and spine cord and nerves

4.5 Architectural roles of the structure based on the human spine

By understanding the anatomical roles of the human spine and getting the building structure based on, there are some capacities, which are added to the structure by building in this way.

1- Due to being this structure as a module, there is a capacity of scaling this structure, so the cortical part as a main structure of the building is scaled (Diagram 4-1).

Diag 4-1: The capacity of scaling the structure based on the anatomical role of human spine

2- Due to being this structure modular, there is a capacity of repeating this structure in different scale as the top of each other (Diagram 4-2).

Diag 4-2: The capacity of repeating the structure based on the anatomical roles of the human spine

3- As this structure is made based on the human spine, it can do its functions and roles, so there is a capacity of making structure in the roles of leaning, crawling, jumping, and etc. for making the different structures such as high-rise buildings, bridges, and etc. (Diagram 4-3).

Diag 4-3: The capacity of making structures in different form of human body

4- The human spine has a stable form while moving, so it is resistance against the internal and external loads. By making structure based on this form, there is a capacity of making a structure, which is resistance against the wind and earthquake loads (Diagram 4-4).

Diag 4-4: The capacity of making structure, which is resistance against the wind or earthquake

5- The human structure has a light structure because it's percent of the weight according to its shape is less. Therefore, by making structure in this way, there is a capacity of decreasing the weight amount of the buildings making the light structure (Diagram 4-5).

Diag 4-5: The capacity of making the light structure

6- The spinal cord is crossed in the middle of the vertebrae of the human spine, so there is a capacity of having crocking way between the parts of the structure, which is made, based on the human spine (Diagram 4-6).

Diag 4-6: The capacity of putting the crossing way inside of this structure based on the anatomical roles of human spine

7- In the middle of the vertebrae is spongy and it doesn't have special roles in loadings bars, if there isn't it, the structure has the empty role inside. So if the structure is built in this form, there aren't any beams inside it (Diagram 4-7).

Diag 4-7: The capacity of making structure without having the beams inside of its structure

4.6 Structures of the other animals spine

The general structure of vertebrae in other animals is the same as human's vertebra, but there are some differences at their ends:

- Acoelous: central with flat ends, like mammal's vertebra (Picture 4-30a), good for supporting and distributing compressive forces.
- Amphicoelous: central with both concave ends, like fish vertebra (Picture 4-30b), the most motion is limited.
- Procoelous: anteriorly concave and posteriorly convex. Like frogs (Picture 4-31a) and modern reptile's vertebra.

- Opisthocoelous: the opposite, possessing anterior convexity and posterior concavity, like salamanders and in some non-avian dinosaurs vertebra (Picture 4-31b).

Pic 4-30: (a) The mammal's vertebrae, (b) The fish vertebrae

- Heterocoelous: saddle-shaped, like turtles (Picture 4-31c) and birds, it makes extensive lateral and vertical flexion motion without stretching the nerve cord too extensively or wringing it about its long axis.

Pic 4-31: (a) Frog vertebra, (b) The non-avian dinosaur's vertebra, (c) The turtle's vertebra

4.7 Preliminary conclusions

The spine which is the biggest and the most important structure of the human body system has a simple structure, but its functioning is very various and complex. In spite of having behaviors and displacements in all directions, its structure would not collapse and is always trying to keep its center of gravity. It is possible to create very different structures with different capabilities by simulating and changing it which has been neglected by designers and architects so far (Salsabili, 2014).

According to what is described, the spine consists of some main levels that include 1- joints (vertebrae, discs, and facet joints) 2- ligaments, 3- muscles, and 4- spinal cords and nerves which are worked together to make both stable and moveable structures (Salsabili, 2014).

In summary, vertebrae include a trabecular system similar to those of present architectural structures to transmit pressures or loads by them. Discs, which consist of nucleus and annulus fibers, are like air or silicone mattresses that their task is to transfer the load from one vertebra to another. Vertebral ligaments are like cables that hold vertebrae together in different directions. They only have a tensile role and prevent the dislocation of the vertebrae from their place. Muscles are like messaging engines, which move and displace the whole structure in different directions. The spine cords are the same as a telephone line to transfer data from

the center data to each aim (Picture 4-32) (Salsabili, 2014). So, there is capacity of making the structure based on the human spine.

Pic 4-32: The structure based on the main part of human spine

By recognizing the spine and its curves, it is possible to simulate different forms and states of the human body (lying, sitting, jumping, bending, etc.) at different ages (infancy, childhood, youth, aging, weightlifting, field, and tracks, etc.) with different curves (Salsabili, 2014).

CHAPTER V: THE MECHANICAL ROLES OF THE HUMAN SPINE

5.1 Mechanical properties of the human spine

In this chapter we will analyze the implications and application of human spine to make the better structure based on the human spine. The main and important part of making structure based on the human spine is knowing the mechanical behavior of each part of the human spine for modeling, simplifying and analyzing for validating the new structure.

5.1.1 Mechanical properties definition

- **Stress:** the amount of forces exerted on an object, divided into a cross-section of the material (Formula 5-1). So, we can use the same yield stress for the same material, without the attention of their size (Parsons, n.d.). The standard unit of measuring the stress: pounds per square inch or Newton's per square meter (Redmond, n.d.).

$$\sigma = F/A$$

Where: σ is Stress

(5-1)

F is force

A is cross sectional area

- **Strain:** the amount of deformation of an object (elongate or shorten) under applied forces. For computing, the changes in length are being divided into object's original length (Formula 5-2) (Parsons, n.d.). There is no unit for measuring the strain because it is a ratio of the deformation over the initial length. The strain measured at 0.05 means that 0.05 inches of deformation for every inch of length (Redmond, n.d.).

- **The Stress-Strain Curve:** stress and strain are directly related to each other; as one increases the other increases as well. So, the more stress, the more deformation of that until failing the object. The Stress-Strain Curve (Figure 5 - 1) is for visualizing the difference in the material properties (Parsons, n.d.). This curve contains three regions: The primary linear part (modulus), the post-yield region (strength properties), and the yield part (the junction of the post-yield and linear parts) (Keaveny & Hayes, 1993).

Fig 5-1: Stress-strain curve

- **Elastic deformation:** the kinds of deformation that will automatically reverse itself after removing external forces. Elastic deformation is linear. Its slope of the line depends on the material of the object.

- **Plastic deformation:** it is permanent deformation. So, it needs an additional external force to return to its original shape. Plastic deformation is not linear, so it is too difficult to model (Parsons, n.d.).

- All objects at first will begin experiencing elastic deformation, and after receiving stress on a certain amount, it will change to plastic deformation.

- **Viscous materials** (fluid material) have fluid's resistance to flow, so their strains are linear with a time of applying stress (Meyers & Chawla, 2008) and absorbing heat during it. But their viscosity reduces by heating ("The Difference Between Elastic Materials and Viscoelastic Materials," n.d.).

- **Viscoelastic behavior** exhibit both viscous and elastic character in deformation. So they are time-dependent material and absorbing heat (Meyers & Chawla, 2008). So they are used for absorbing shock, dampening noise, and isolating vibration ("The Difference Between Elastic Materials and Viscoelastic Materials," n.d.).

- **Yield Stress:** the point of changing elastic to plastic deformation. Computing the yield stress of an object determines experimentally (Figure 5-2a, 5-2b) (Parsons, n.d.).

- **Yield point (strength):** some material (steel material) has a yield point (Figure 5-2c), at this point, there is a sharp increase in the object's strain without correlation with an increase in stress. It happens after reaching its yield stress (Parsons, n.d.).

Fig 5-2: (a) Load depends on Yield points, (b) Yield stress point, (c) Yield point

- **Brittle materials** (glass) will break or fracture without bending when a large enough force is applied. So, their yield point is the same as the fracture point (Figure 5-3a) (Redmond, n.d.).

- **Ductile materials** (steel or aluminum): if the force is large enough, the material will permanently deform and bend so they cannot return to its original shape. This dip point in the curve indicates that the material has yielded or deformed (Figure 5-3b) (Redmond, n.d.).

Fig 5-3: (a) Brittle material, (b) Ductile material

- **Mechanical testing:** this test involves applying compressive, tensile, shear, torsional loads to record the stress-strain curves (material properties) and Force-deformation (structural properties) (Cat, 2010).

- **Absorbed energy (toughness):** the area under The Stress-Strain Curve before failing shows the absorbed energy. So it depends on the ultimate strain and force. It shows the situations of absorbing energy by bone in a force situation (Keaveny & Hayes, 1993).

- **Modulus of elasticity (stiffness):** it is the ratio of stress-strain or a tangent of the stress-strain curves. It uses to measure the stiffness of a material, not elasticity (Gaynor, 1969). For example, oak, stainless steel, and polyvinyl chloride (PVC) have Young's modulus of 200 GPa, but trabecular bone and whole bone have approximately Young's modulus of 1 and 17 GPa, respectively. So the stainless steel is over ten times stiffer than bone (Keaveny & Hayes, 1993).

- **Density:** the ratio between the mass of mineralized bone tissue to the bulk volume of the mineralized bone plus bone marrow spaces of that specimen (Keaveny & Hayes, 1993).

- **Poisson's ratio** is the ratio between the longitudinal strain and transverse strain ("Strain, Stress, and Poisson's Ratio," n.d.). It describes how much a material bulges in compression, or contracts in stretching (Formula 5-3) (Typically 0.3 for metals). Usually, when materials are stretched in one direction they tend to get thinner in the lateral direction and in compression they get the opposite role (Gercek, 2007). Except for auxetic materials like some polymer foams or origami folds that have stretched or compressed in both direction at the same time (Mousanezhad et al., 2015), Most of the materials have Poisson's ratio ranging between 0.0 and 0.5 (R. J. T. Park, 1987).

- **Creep Behavior:** the deformation of bone tissue in stable stress for a long period of the time (Keaveny & Hayes, 1993).

- **Isotropic materials** (like 316L stainless steels): elastic properties of them respect to the Poisson's ratio and Young's modulus, but not depend on the orientation of the material (Keaveny & Hayes, 1993).

- **Anisotropic materials** (like bones): they are materials with directional-dependent elastic properties. These five parameters are necessary for them: the longitudinal moduli, the transverse moduli, two Poisson's ratios, and the shear modulus (Keaveny & Hayes, 1993).

- **Fatigue:** It may happen with the high levels of stress over shorter periods of time (Keaveny & Hayes, 1993).

- **Force** is anything, which tends to change the state of a body (producing stress and strain within the body) such as: compressive (pushing together), tensile (pulling apart) and shearing (forces of sliding one part with an adjacent part) (Figure 5-4).

Fig 5-4: Types of force

- **Mechanics**, the science dealing with the effect of forces upon the form or the motion of bodies, has two types:

- Statics is the study of bodies at rest or in equilibrium. Mechanical properties of materials are usually studied under static conditions under a slowly applied force or load to be easier to analyze.
- Dynamics is the study of moving bodies (Gaynor, 1969).

- **The Structural properties** are generally described by four parameters including (Figure 5-5a):

- Stiffness (slope of the linear portion of the load-elongation curve).
- Ultimate load (maximum load at which the complex fails).
- Ultimate elongation (elongation corresponding to with the maximum load).
- Energy absorbed in failure (area under the curve to the maximum load) (Parsons, n.d.).

- **The Mechanical properties** are generally described by four parameters including (Figure 5-5b):

- Tangent modulus (slope of the linear portion of the stress-strain curve).
- Tensile strength (stress at failure).
- Ultimate strain (strain corresponding to the tensile strength).
- Strain energy density (area under the stress-strain curve until failure) (Parsons, n.d.).

Fig 5-5: (a) The structural properties, (b) The mechanical properties

- Stages of analyzing:

- The **material behavior** (opposed to structural properties): determining of tissues which makes up the whole bone that they are independent of the geometry of the bone from which the tissue is obtained (Keaveny & Hayes, 1993).
- The **geometric properties**: determining the whole bone form and structure (Keaveny & Hayes, 1993).
- The **structural analysis**: performing data based on the combined effects of material and geometric.

5.1.2 Bone

5.1.2.1 The samples of studying

As far as I am aware, there aren't enough experimental studies on the mechanical properties of living human bone for obvious reason. But, there is an experimental investigation of the mechanical properties of dead and living bones of rats (Evans, 1973). After an experimental investigation, the dead bone retained its original radiographic density, specific gravity, strength, and composition. Because there are no significant differences in bone viscoelastic, anisotropic, composite material (organic fibers (collagen), inorganic crystals (hydroxyapatite), cement substance, and water). So, it is valid for assessing similar properties of living bone (Evans, 1973).

5.1.2.2 Loading the bar of bones

Bars on bones come from external forces which are applied to the body and internal forces that are generated by the connective tissue, muscle, and muscle forces (Cat, 2010).

5.1.2.3 Composition of bones

The bone is composed of four main parts such as:

- **Mineralized collagen fibrils**, at the nanometer scale ($\approx 0.1\text{-}\mu\text{m}$ scale). Collagen fibrils which have the size from 20 to 40 nm in diameter and number from 200 to 800 collagen molecules in their cross sections of a fibril (Keaveny, Morgan, & Yeh, 2004).

- On a weight basis: bone is approximately 10% water, 60% inorganic and 30% organic.

- On a volume basis, bone is about 25% water, 40% inorganic and 35% organic.
- The inorganic phase (hydroxyapatite): this impure mineral consist of crystals such as potassium, magnesium, strontium, and sodium, carbonate, and chloride or fluoride which are made the bone hard (Keaveny et al., 2004).
- The organic phase: collagen type I (90 percent by weight), collagen types VI and III and a different types of noncollagenous proteins (Osteocalcin, Osteopontin, Osteonectin, and Bone sialoprotein) which are arranged in parallel with a hole zone (gap) of 40 nm between each molecule to make the bone flexibility (Cat, 2010).
- **Lamellae woven**, at (1- to 10- μ m scale), they arrange in two types of forms:
 - Lamellae (stacked of thin sheets) have 7 μ m thickness that consists of unidirectional fibrils in intermittent angles between their layers. They are common in adult human's bones.
 - Woven fibrils (with the forms of block randomly oriented fibrils), woven bones happen in situations of the rapid growth of children and large animals (Keaveny et al., 2004).
- **Haversian Lamellae weaved (Osteonal)**, the most usual type of cortical bone in adult humans bones in which about 10 to 15 lamellae are arranged around a central Haversian canal with a channel (50 μ m) of nerves, a variety of bone cells, and blood vessel capillaries (Keaveny et al., 2004).

Pic 5-1: Bone component

- **Cortical and trabecular** (the fourth level of the bone microstructure (1 to 5 mm)) (Picture 5-1).
 - Cortical (a tightly packed with lamellar, haversian, or woven bone).
 - Trabecular (a highly porous cellular solid (open-celled)) bone.
 - Porosity (The distinction between cortical and trabecular bone).

Generally, all bone is equal in their materials, but they may have little difference in their lamellar architecture or degrees of their mineralization between trabecular and cortical bone (Keaveny et al., 2004).

5.1.2.4 Trabecular and cortical bone

Cortical and trabecular bone have differences in their porosity, density and in their bone architecture (Figure 5-6) (Keaveny & Hayes, 1993). Cortical bone is distinguished by its low porosity and osteonal microstructure, while trabecular bone is distinguished by its high porosity and open-celled architecture.

Fig 5-6: The trabecular and cortical

5.1.2.4.1 Mechanical properties of cortical bone

- Cortical bone is stronger and stiffer in “transverse” directions when loaded longitudinally (Keaveny et al., 2004).

Fig 5-7: The stress- strain curve of bone

- Cortical bone is stronger in compression than in tension (Figure 5-7). The strength-to-modulus ratio for cortical bone is about 0.78 for longitudinal tension and 1.12 in longitudinal compression (Aluminum 6061-T6:0.45 /titanium 6A1-4V: 0.73). So cortical has a high-performance, particularly in compression (Table 5-1) (Keaveny et al., 2004).

Table 5-1: The Ultimate stress of human femoral cortical bone

Ultimate stresses of human femoral cortical bone			
Longitudinal (MPa)		Transverse (MPa)	
Tension	135(15.6)	Tension	53(10.7)
compression	205(17.3)	compression	131(20.7)
-		Shear(MPa)	65(4.0)

- Cortical bone has a linearly elastic material behavior and fails after its yield point with small strains (0.2 percent) and without having any plasticity. Cortical bone longitudinal tensile ultimate strains are up to 5 percent in young adults then decrease to 1 percent in the elderly ages (Table 5-2).

Table 5-2: Elastic properties of human femoral cortical bone

Elastic properties of humn femoral cortiacal bone	
Longitudinal modulus (MPa)	17900 (3900)
Transverse modulus(MPa)	10100 (2400)
Shear modulus(MPa)	3300 (400)
Longitute poissons ratio	0.40 (0.16)
Transverse poissons ratio	0.62 (0.26)

- Cortical bone is so weak in the sheer situation and is getting weakest in tension when is loaded transversely (Keaveny et al., 2004).

- The creep rate increases by increasing stress, and face a larger permanent deformation after it's unloading.

- Cortical bone has a viscoelastic behavior, but its loading rate on modulus and strength has a moderate effect (Table 5-3).

Table 5-3: Strength properties of human femoral cortical bone

Strength properties of human femoral cortical bone			
Longitudinal (MPa)		Transverse (MPa)	
Tension	133	Tension	51
compression	193	compression	133
-		Shear(MPa)	68

- Aging decreases the tensile ultimate stress (approximately 2 percent per decade) and it more happens in women (Figure 5-8a).

- Aging makes both the modulus and strength properties of cortical bone worse with the similar rate for men and women (2%/decade after age 20) (Figure 5-8b) (Keaveny & Hayes, 1993).

- Energy absorption decreases 7% per decade because of its reductions in the ultimate strain. So, with aging cortical bone material becomes less strong and more brittle (Keaveny & Hayes, 1993).

Fig 5-8: (a) Strain rate by aging, (b) Strength amount by aging

- We face the reduction when cortical bone is loaded after its yield point, unloaded, and reloaded again. We aren't seen this for metals, their reloading modulus is the same as the primary modules (Figure 5-9) (Keaveny et al., 2004).

Fig 5-9: Stress-strain curve in loading, unloading, and reloading

-. Cortical bone is tough material and their fatigue strain is negligible, because of its resistance to fatigue failure in compression than in tension (Table 5-4) (Keaveny et al., 2004).

Table 5-4: Elastic properties of human femoral cortical bone

Elastic properties of human femoral cortical bone	
Longitudinal modulus (MPa)	17000
Transverse modulus(MPa)	11500
Shear modulus(MPa)	3300
Longitude poissons ratio	0.46
Transverse poissons ratio	0.58

- Cortical bone is an anisotropic material because it's elastic properties dependent on direction, but cortical bone is an isotropic material in each plane (Keaveny & Hayes, 1993).
- The longitudinal direction modulus of cortical bone is 1.5 times larger than its transverse direction modulus and over five times of its shear modulus.
- The Poisson's ratio of cortical bone (up to 0.6) is high, it shows that its bulges more than metals in compression.
- Cortical bone depends on the loading direction and whether compression or tension forces are applied can behave as a ductile or brittle fashion (Keaveny & Hayes, 1993).
- Strain rate increases as the activity becomes more intense. So, the cortical bone becomes stiffer, stronger, and can absorb more energy as the loading rate is increased (Figure 5-10) (Keaveny & Hayes, 1993).
- Density of cortical (mean values of 1.85 g/cm³) does not depend on anatomic sites or species, but it is sensitive to aging (Keaveny & Hayes, 1993).

Fig 5-10: (a) Strain- time curve, (b) Density of cortical

5.1.2.4.2 Mechanical properties of trabecular bone

- Trabecular bone (cancellous or spongy bone) has a nonlinearly elastic effect even for small strains, but usually is modeling as linearly elastic before its yielding point (Figure 5- 11a).

Fig 5-11: (a) Stress curve of trabecular by aging, (b) Yield stress of trabecular bone in different directions

- Trabecular bone yields in compression are at strains of approximately 1 percent and after that yield point; it faces large deformations (up to 50 percent strain). So, it can absorb significant energy from mechanical failure and its failure (yield and ultimate) strains have a weak dependence on modulus and density (Figure 5- 11b).
- Trabecular bone has the modulus in the range of 10 to 3000 MPa depends on the past factors.
- Trabecular bone strength is linearly and strongly correlated with its modulus (in the range of 0.1 to 30 MPa).
- Both modulus and strength (in the range 0.5 to 0.9) depend heavily on apparent density, age, anatomic site, disease, (Keaveny et al., 2004) and loading mode (being lowest in shear, highest in compression (Figure 5-12a)) .
- The strength of trabecular bone is weakened by a decrease in thickness, reduction in number, and increase in length (Keaveny & Hayes, 1993). So we face finally reductions in vertebral failure load (Figure 5-12b) (Silva & Gibson, 1997).

Fig 5-12: (a) Stress-Strain curve of trabecular, (b) Yield strain in tension or compression

- For example, stress strength decreases by approximately 10 percent per decade by aging (Figure 5-13).
- Yield stains are less in tension than compression (1.0 to 2.5 percent).
- Tensile yield strains are less than compressive yield strains too.
- Poisson's ratios with a range of 0.03 to 0.60 by experimentally are difficult to measure (Keaveny et al., 2004).

Fig 5-13: Strength and density by aging

- Trabecular bone is loaded, unloaded, and reloaded in compression in its elastic range, displays a decrease in stiffness and increase of permanent strains (Figure 5-14) (Keaveny et al., 2004).

Fig 5-14: Stress- strain curve of loading, unloading a reloading of trabecular bone

- Human trabecular bone is nonlinearly viscoelastic (Keaveny et al., 2004).
- Trabecular bone has an isotropic role in some anatomic sites and highly anisotropic in others (elderly lumbar spine) in contrast to cortical bone. So, trabecular bone is usually supposed to be isotropic in whole-bone (Keaveny et al., 2004).
- Trabecular tissue has an elastic behavior (its modulus is between 1 to 10 GPa, 20 percent lower than its cortical tissue elastic behavior) (Keaveny et al., 2004).
- Yield strains of trabecular tissue the same as cortical bone is less in tension than in compression.
- The fatigue strength of cortical tissue is higher than of its trabecular (Keaveny & Hayes, 1993).
- The field of bone cortical and trabecular mechanics are sophisticated because of different anatomic sites, aging, bone density, and disease on their properties (Keaveny et al., 2004).
- The average density of trabecular bone is depended on its anatomic site (0.10 g/cm³ for the spine, 0.60 g/cm³ for the femur, and 0.30 g/cm³ for the human tibia) (Keaveny et al., 2004).

- The average density of trabecular bone is depended on ages. After the maturity of skeletal (around 25 to 30 ages), its density decreases steadily at a rate of 6 percent per decade (Picture 5-2) (Keaveny et al., 2004).
- The thicknesses of individual trabecular is in the range 100 to 300 μm interspace of 500 to 1500 μm between two trabecular (Keaveny et al., 2004).
- The material properties of trabecular bone depend on apparent density, the architecture of the trabecular bone, the age of the tissue, and orientation of its basic cellular structures.
- The longitudinal Young's modulus and hardness of human trabecular bone are 20.78 ± 2.4 GPA with the Poisson's ratio of 0.3 (Giambini et al., 2013).
- It is the compressive stress-strain behavior of cortical and trabecular for different densities. They have the yields point at strains in the range of 1 to 4% without dependence on density (Figure 5-15) (Keaveny & Hayes, 1993).

Pic 5-2: different density of trabecular bone

Fig 5-15: Stress-strain curve of trabecular and cortical bones

- Tensile stress-strain behavior for trabecular bone shows that failure happens by fracture of the individual trabecular. This behavior is similar to that for fiber-reinforced concrete and it is typical of artificial materials, which are designed to resist compressive forces (Figure 5-16a).
- The post-yield load-carrying capacity of trabecular bone is high for compression and almost negligible for tension (Keaveny & Hayes, 1993).
- It is interesting that the strength and modulus adjust equally to changes in strain rate the same as changes in density (Keaveny & Hayes, 1993).
- The modulus and strength are significantly higher for specimens with bone marrow (Figure 5-16b) (Keaveny & Hayes, 1993).

Fig 5-16: (a) Stress- strain curve after fracture of trabecular, (b) Stress- strain curve of trabecular with different marrow

- Losing bone mass causes density reduction and changing architecture. This reduction depends on gender, anatomic site, and age (reduction of 50% from ages 20 to 80 years) (Picture 5-3).
- The thickness and number of the trabeculae bone decrease with decreasing of its density but we face an increase in the size of the intertrabecular (Keaveny & Hayes, 1993).
- The number of horizontal trabecular is less than the number of vertical trabecular, regardless of density, and decreasing the density of vertical trabecular is twice the rate of horizontal trabecular (Keaveny & Hayes, 1993).
- Vertical trabeculae do not become thicker with aging, and preferential loss of horizontal trabeculae does not appear to occur in the lumbar spine. In fact, these three-dimensional data suggest that preferential loss of vertical trabecular occurs with increasing age and loss of trabecular is damageable than being trabecular thinner (Figure 5-17) (Keaveny & Hayes, 1993).

Pic 5-3: stress- strain curve of trabecular with different marrow

Fig 5-17: The number of trabecular in vertical and horizontal

5.1.2.5 Factors affecting the mechanical behavior

- Disease

For example, osteoporosis is a kind of disease with reduction silently and progressively in density and quality. So the risk of fracture is greatly increased (Picture 5-4) (“What is Osteoporosis?,” n.d.).

- Elastic loading is the same as normal bone without a reduction in yield stress and elastic energy absorption in this disease (Dickenson, Hutton, & Stott, 1981).

- Facing the reduction of energy absorbing before failure in the plastic region of bone shows that the bone is more sensitive to fracture from injury (Figure 5-18).

Pic 5-4: healthy and osteoporosis bones

Fig 5-18: Stress- strain curve of healthy and osteoporosis bones

- Drying

The freshness of the bone tissue influences the properties of bone (Cortical bone strain is 1.2 % when it is wet and 0.4% in dryness) (Cat, 2010).

The ultimate tensile strength, modulus of elasticity in tension, hardness and stiffness (Gaynor, 1969) are increased by drying while the tensile strain and the fatigue life of the bones are reduced (Evans, 1973).

Wet bone is also less strong, less stiff, and less brittle than dried bone (Figure 5-19a) (Cat, 2010).

- Embalming

Because of the difficulty of securing a sufficient amount of fresh on embalmed material made embalmed bones. But there are some effects on the mechanical properties of bone (Evans, 1973). Embalmed cortical bone have a more compressive strength, the more modulus of elasticity, the less energy-absorbing capacity and the less strain than the normal one (Figure 5-19b) (Gaynor, 1969).

Fig 5-19: (a) The mechanical behavior of dry and wet bones, (b) Mechanical behavior of embalmed and normal bone

- Age

We face increased of skeletal mass during the growth period rapidly (25 to 30 years) and reduction throughout the rest of life slowly. Mean modulus of elasticity (from 17 GPA to 15.6 GPA), hardness steadily, the tensile strength (from 130 MPa to 110 MPa), the energy absorbing capacity of bone (7% per decade) (Gaynor, 1969) decrease by aging (Evans, 1973), but we have increased in the slope of the stress-strain curve after yielding 8% per decade (Figure 5-20). The cortical bone material becomes less stiff, less strong, and more brittle with aging (Cat, 2010). The trabecular bone face decreased in the thickness of its network (horizontal trabecular not vertical) and an increased in the distance between them with aging (Giambini et al., 2013).

Fig 5-20: Mechanical behavior by aging

- Gender

The mean ultimate tensile strength and modulus of elasticity are greater for men than women. But mean tensile strain is greater for women (Figure 5-21a) (Evans, 1973).

- Race

The mean ultimate tensile strength is greater in white males than in black males (Evans, 1973), and the mean tensile strength is greater in white women than black men (Figure 5-21b).

Fig 5-21: (a) Mechanical behavior by gender, (b) Mechanical behavior by race

5.1.2.6 Mechanical behavior of whole bone

- Bone is the hardest materials of the body with the capacity of repairing itself and adapting environmental stresses changes. The turnover rate for cortical bone is 50 % annually in children, 10% for adult and 2 % in the elderly. The turnover of cortical is 5 to 10 times less than trabecular bone because of its less surface area (Cat, 2010).
- Bone has a strain rate sensitive than other biological tissues because of its unfixed porosity in the response to loading, disease, race, gender, and aging. This cause for bone-ligament and bone-tendon to injury (Figure 5-22a) (Cat, 2010).
- Both glass and steel are stronger and stiffer than bone (Figure 5-22b).

Fig 5-22: (a) Stress- strain curve of whole bone, (b) Stress-strain curve of glass, metal, and bone

- The optimal strain rate of energy absorption happens between 0.1 and 1/second (Figure 5-23a) (Cat, 2010).
- The maximum load before failure depends on the loading rate and the type of load applied (tensile, shear, and compressive) (Cat, 2010).

- The deformation of failure depends on the loading rate and its direction (Figure 5-23b) (Cat, 2010).

Fig 5-23: (a) Stress- strain curve by load/sec, (b) Different type of bone loadings

- Children's bones tend to have more energy absorbing before failure than adults (45 % more).
- Children's bones are compliant but weak. (68% stiffness of adult bone) (Cat, 2010).
- The mechanical properties of the trabecular tissue are lower than cortical bone, but Carter and Hayes said that the trabecular had the same mechanical properties as a cortical bone (Young's modulus of 20 GPa) (Nicholson et al., 1997).
- The whole vertebrae have the same elastic modulus and hardness, but the wet vertebrae have 25% reduction of moduli and ~57% of hardness than a dry one (Giambini et al., 2013). So, highly mineralized bone have higher stiffness and compressive strength than the less mineralized one.
- The modulus of trabecular bone (10 to 2000 MPa) depends on the anatomic site and age. But the modulus of cortical bone is approximately 17000 MPa, this indicates that trabecular bone is much more compliant than cortical bone (Keaveny & Hayes, 1993).
- There aren't any differences between the posterior and anterior regions in Young's modulus values and hardness ($P < 0.05$) (Figure 5-24a) (Giambini et al., 2013).
- Mean and range of variation in the strength of various bones and direction of loading are shown in the figure (Figure 5-24b) (Gaynor, 1969).
- The fatigue life of bone is better correlated with strains than stresses. The fatigue life of bone is also dependent on temperature (Keaveny & Hayes, 1993).
- Cortical bone is stiffer, less strain before failure (fracture exceeds 2 %), face greater stress, less energy-absorbing and less absorbed energy than trabecular (Cat, 2010).

Fig 5-24: (a) Hardness and young's modulus of interior and posterior bone, (b) The strength of various bones

5.1.3 Intervertebral disc

5.1.3.1 The composition of intervertebral disc

Intervertebral discs are deformable therefore they allow small movements between adjacent vertebrae so the whole spine can bend 6° to each vertebra, twist and bear the weight by distributing the mechanical loading along the column.

The disc includes three anatomic zones: the central nucleus pulposus (NP), the outer annulus fibrosus, and the intermediate transition zone (Jongeneelen, Baaijens, & Huyghe, 2006).

- **Nucleus pulposus (NP):** the soft deformable central region of the disc is contained of a gelatinous, a proteoglycan-water gel (the large molecules and sugar subunits with having a strong electrostatic attraction of water), a mesh of fine collagen fiber with a randomly oriented network (type-II with diameters of 30–500 nm), and non-collagenous proteins. The collagen network is in the interconnection with the endplate and the inner annulus. The high concentration of proteoglycan attracts water (70 to 90 % of nucleus content) (Jongeneelen et al., 2006) and gives the nucleus the capacity of swelling (highest in night-time rest, and lowest in daytime activity) (Adams, 2015). Nucleus primarily contributes to the torsional or twisting movement of the body (Singha & Singha, 2012).

Endplate is part of the vertebra (approximately 0.5 mm) which is composed of hyaline cartilage (similar chemical composition of the nucleus with collagen type II to make it rigid and prevents swelling) plays a vital role in disc mechanics. The incompressible fluid nucleus presses on the bony endplate and makes bulging up to 1 mm into the vertebral body under compressive loading (Adams, 2015). These fluids flow inside the end plate play the main role in the recovery of the disc (Singha & Singha, 2012). The cartilage of endplate is thick adjacent to the nucleus and inner annulus and it is absent in the outer

annulus. It has a weak bonded to the cortical bone of the adjacent vertebral bodies (Adams, 2015).

Pic 5-5: Intervertebral disk

Fig 5-25: Intervertebral annulus

- Annulus Brosius: the outer boundary of the disc consists of two regions: the inner transitional area adjacent to the nucleus and the outer collagenous area (Picture 5-5). From it's outer to inner regions face changing of composition gradually (Jongeneelen et al., 2006).

Annulus contain highly oriented type I collagen fibers (20–60) in parallel bundles to make lamellae (15–25 concentric) with angles of 40° and 70° to the vertical axis around the nucleus, which are surrounded by a proteoglycan gel (Figure 5-25). The annulus has a crimped planner (zig-zag) structure, so it enables them to stretch more, poroelastic mechanical profile, viscoelasticity, aggregate or elastic modulus, anisotropic or heterogenetic, permeability and disc tissue porosity, and governs four main biomechanical spinal disc manifestations (stress-strain rate trend, creep, hysteresis and distribute the stress on spine) (Adams, 2015; Singha & Singha, 2012). It is much stronger than NP so degeneration of annulus mainly causes LBP (lower back pain). The AF layers have a special arrangement of fibers; therefore, they face more load-bearing capacity (mainly in compression, shear, torsion, tensile etc.) (Figure 5-26). The Poisson's ratio of AF is higher than the NP; therefore, AF has a higher degree of deformation. it shows

Fig 5-26: Annulus direction of loadings

Pic 5-6: Annulus Lamellae

aging and degenerative changes sooner than any other connective tissue in the human body (Jongeneelen et al., 2006).

Adjacent lamellae (Picture 5-6) (Pezowicz, 2010) are bonded together loosely, a fibrous protein in annulus and nucleus are elastin so can be stretched vertically without damage (with strains of 10 % (similar to the tendon)). In large deformations, the shape recovered by changing the compressive force to lateral force and tolerate external tensile stresses (circumferential, torsion, and longitudinal) (Singha & Singha, 2012). This elastin network is in the nucleus and between the adjacent lamellae of the annulus (Figure 5-27) (Adams, 2015). In the inner zone, annulus are crimped to cartilaginous endplates while in the nucleus and in the outer zone crimped to the vertebral bodies (Jongeneelen et al., 2006). The cartilage endplate is a dense filter by reducing the rate of compression by expelling the water from the nucleus into the vertebral bodies to sustained loading (Adams, 2015).

- Transition zone (inner annulus): the inner annulus usually are named as the Transition zone (Adams, 2015).

Fig 5-27: Annulus directions of loadings

5.1.3.2 The intervertebral disc components

The structure, organization and molecular interaction of proteoglycan, collagen, glycoproteins (density of 9000 cells/ mm³ in the annulus and about 4000 cells/mm³ in the nucleus) and water make the functional properties of the disc (Figure 5-28) (Jongeneelen et al., 2006).

- Collagen (similar to articular cartilage): the content of collagen decreases from the outer of the annulus to the center of the nucleus pulposus. Collagen provides a hard network to support the cells and connect the hydrated proteoglycans (Jongeneelen et al., 2006).

- Proteoglycan: the major structural components of the intervertebral disc (Approximately 30% dry weight of IVD) to influence its mechanical properties. By attracting water, it brings the swelling capacity to the nucleus (Osmotic pressure) (Jongeneelen et al., 2006).

- Cells (Only 1% of the disc volume) make three types' tissue components such as the chondrocyte-like (nucleus), fibrocartilaginous cells (the inner annulus) and fibroblast-like cells (the outer annulus). The biomechanical behavior of the disc depends on the composition of these discs cells (Jongeneelen et al., 2006).

- Microenvironment: the roles of mechanical environment of chondrocytes and having conjunction with other genetic and environmental factors plays an important role in the health of the joint and cartilage homeostasis (Jongeneelen et al., 2006).

- Water: 60- 70% of the IVD is water. In the depth of the disc, there is a variation in a component of PG (Proteoglycan) and water. In the upper subchondral bone, the concentration of PG is less and water is high, but inside the tissue, the concentration of the PG is high and water is less (Singha & Singha, 2012).

Fig 5-28: The intervertebral components

So, there is increasing hydration (from 60% to 80%), proteoglycan content, collagen type I content (from 100% to 0%) from outer to inner of annulus to increase density (Jongeneelen et al., 2006) (Adams, 2015).

Table 5-5: The component of IVD

Components	AF	NP
Water	60-70%, no charge with age	90% at birth
		80% at age 20
		70% of older age
Collagens (collagen, collagen, collagen - collagen is produced by the degenerated disc which has very poor mechanical properties)	Only collagen, %0-60% with (dry weight)	Only collagen .15-20% with (dry weight)
	Little change with age	Little change with age
PGs (Proteoglycans)	15-20% with (dry weight)	65% with (dry weight)
	Little change with age	Little change with age
Non- collagenous proteins and elastin	5-25% with (dry weight)	20- 45% with (dry weight)
	Little change by age	Little change with age
Extracellular enzymes, age pigments, cells	Minor remainder	Minor remainder

For simulating the intervertebral disc structure we have to pay attention to the nutrient supply, the mechanical stress, and the ionic and osmotic environment (Table 5-5) (Jongeneelen et al., 2006).

5.1.3.3 Mechanical behavior of the intervertebral disc

The disc task is making a motion in the spine, loading bars (static loads, dynamic loads, torsional loads, tensile stresses, shear stresses and a combination of tensile, compressive and shear stresses), and maintaining flexibility (Jongeneelen et al., 2006).

- The annulus (AF) because of its lamellar networks of collagen fibers provides stiffness for the disc to support mechanical loads. The nucleus because of its high water content increases the pressure on the disc. The endplate is porous and allows fluid flow through the wall of the annulus between the disc and vertebral body (Gabai, 2007).

- The nucleus (NP) task is loading the compressive and the annulus task is loading the tensile stresses.

- Cells in the nucleus are stiffer than in the annulus. Nucleus has small compressive and tensile strains because of its large hydrostatic pressures and little volume change. In contrast, cells of the lamella in the annulus are well adapted for small tensile strains and volume changes in tensile loading and of course having higher compressive strains (Jongeneelen et al., 2006).

- Outer of the annulus is more solid to acts highly in tension and poorly in compressive. In the middle annulus, the stress concentrations increase especially in the posterior of the nucleus (transition zone). In nucleus pulposus, vertical and horizontal stresses are equal at each location (resistance with strains up to 30 %) to support high compressive loads daily (Adams, 2015).

- Inner annulus are thicker and have a lower density of collagen fibers in its construction, so it is a transitional structure between the nucleus and the annulus (Pezowicz, 2010).

- Mechanical properties of the component of annulus fibrosis lamella (Table 5-6) (Pezowicz, 2010).

Table 5-6: Mechanical properties of the component of annulus fibrosis lamella

Mechanical properties		Average value (SD)
E_{11} (MPa)	53	53.2 (27.5)
E_{22} (MPa)	6	5.8 (3.12)
V_{11}	0.7	0.66 (0.22)
V_{12}	1.2	1.16 (0.68)
G_{66} (MPa)	0.1	0.11 (0.03)

- The collagen fibers make the tensile strength in tissue and proteoglycans make the tissue stiffness and resistance to compression because of holding the fluid inside the disc (Jongeneelen et al., 2006).

- Collagen “glue maker” (Adams, 2015) in the annulus provides supporting during compression, tensile, bending, torsion of the disc, and preventing tears across the ligaments (Jongeneelen et al., 2006).
- The pericellular microenvironment (PCM) has an influence on the stress-strain, fluid flow and stiffness to become greater than that of the cell (PCM 65 kPa, cells 55 kPa) (Jongeneelen et al., 2006).
- Fluid flow is a major component of disc mechanical behavior due to its role of getting out of the human disc during the day because of mechanical stresses from gravity, muscle contraction, and spinal motions (Gabai, 2007).
- The proteoglycans in the nucleus stabilize the disc by charging glycoproteins to attract water and making the internal swelling pressure to balance external and internal stresses. So the pressure on the nucleus causes fluid to flow out to increase the proteoglycan concentration in the nucleus then the fluid inflow again after removing stress (Gabai, 2007).
- Annulus have the task of swelling only in the vertical (axial) direction (Adams, 2015).
- Disc swelling pressure and its fiber deformation increase linearly by increasing load magnitude between 0.1 and 0.3 MPa and over that become constant (Gabai, 2007).
- Approximately 50 % of body weight and tension of spinal muscles act to make the spine in upright postures due to the spine curves.
- The compressive force acts on the long axis of the spine on intervertebral disc and the shear force acts perpendicular to that (Adams, 2015).
- Spinal can resist backward and lateral bending by the balance coming from the neural arch. The disc with the neural arch can resist an average 33 % of the loading and it can resist 37% of an applied axial torque in backward bending and in axial rotation (Adams, 2015).
- The disc and the posterior intervertebral ligaments can resist 30 % of the forward bending (flexion) and act on the lumbar spine (up to 20 Nm during typical lifting tasks) (Adams, 2015).
- The normal strength of anterior bending is 33 Nm, and failure happens near the junction of the disc bone in the outer of the annulus (Adams, 2015).
- IVD is an anisotropic material due to the different angles in cross-woven of the lamellas (Jongeneelen et al., 2006).
- Intervertebral disc tissues are nonlinear so they are too stiff.
- Intervertebral disc tissues are poroelastic, so based on the rate of loading they have load-sharing (Adams, 2015).
- The component of intervertebral discs and ligaments cause to display viscoelastic behaviors (Adams, 2015).

- The forces apply on AF layer are a tensile force (uniaxial, biaxial, triaxial), a compressive force (uniaxial, biaxial, triaxial), shearing force, twisting force, torsional, and the water hydrostatic force (Singha & Singha, 2012).

-The strains are in the dependency of the structure, behavior of the extracellular component, disc height, permeability, and time-dependent behavior (Gabai, 2007). It does not have to correspond with the applied strains (Figure 5-29) (Jongeneelen et al., 2006).

Fig 5-29: Compressive stress curve

- The strength of 3.9 MPa is for the inner annulus and 8.6 MPa is for the posterior in the vertical direction. So stiffness, strength and tensile properties increase from inner to outer of annulus because of increasing of collagen content (Type I) in the radial direction (prepended to the lamellae) (Table 5-7) (Adams, 2015).

Table 5-7: Tensile properties of annulus fibers

	Inner annulus	Middle annulus	Outer annulus
Failure stress (MPa)	2.0 (1.5)	4.1 (2.3)	7.4 (4.3)
Failure strain (%)	64 (36)	44 (15)	40 (11)
Normalized stiffness	1.2 (1.11)	2.77 (1.62)	5.67 (3.38)
Failure energy (N mm)	15.5 (13.7)	21.6 (15.4)	33.2 (23.9)

- Annulus specimens exhibit stiffer in quick stretching and fatigue failure happens in less than 10,000 cycles of tensile force (Adams, 2015).

- This analysis shows two mechanisms cause damage to annulus fibrosus:

- The first, the tensile stretching over $41.9 \pm 7.9\text{N}$.

- The second, damage in the region of the upper bone-attachment with the average force of $12, 3 \pm 3.4\text{N}$ (Figure 5-30a) (Pezowicz, 2010).

- The linear Young's modulus (the stress-strain characteristics) were divided into three equal ranges (lower, medium, and upper) on a scale from zero to the maximum stress (Figure 5-30b) (Table 5-8) (Pezowicz, 2010).

Fig 5-30: (a) Load- Displacement curve of annulus in tensile (b) Stress- strain curve of annulus

Table 5-8: Young's modulus in the representative range

	E_{low}	E_{medium}	E_{high}
Ist mechanism	39.3+ 5.0	46.9+8.1	35.5+8.0
2 nd mechanism	37.2+3.2	45.7+2.4	31.1+3.6

- Typical stress-strain characteristics, obtained during stretching of an annulus fibrosis lamella (Figure 5-31) (Pezowicz, 2010).

Fig 5-31: Typical stress-strain curve

- For a constant rate of tensile extension (0.4mm/min), all tested samples showed an average stress value of 0.15 ± 0.06 MPa (Pezowicz, 2010).

- Tensile strain depends on orientation and location. The averaged ultimate tensile strain is 63 % (range 26–110 %) and ultimately averaged tensile stress is 0.26 MPa (range 0.08–0.64 MPa) (Adams, 2015).

- During the stretching of micro-samples along collagen fibers, the maximum value of stress coincided with displacement and separation of fibers along their entire length in rapidly declining stress. The connection between borderline cartilage and bone because of its weakness makes this region vulnerable to damage (Pezowicz, 2010).

- In the initial phase, the crimp structure of collagen fibers is progressively straightened, tightened, and faced a reduction in tension results in a return of the crimp nature of collagen fibers. The maximum average values of tension equaled $\max = 16.4 \pm 8\text{MPa}$ (Pezowicz, 2010).

- In unloaded stretching phases, the aligned fibers begin to separate in isolated regions – clefts. With increased stretching, these same clefts open even further (Pezowicz, 2010).

- (a) Annulus fibrosus lamella in the unloaded state (b) separation of collagen fibers during tensile stretching along the fiber alignment direction (Picture 5-7a) (Pezowicz, 2010).

- (a) Annulus fibrosus lamella in the unloaded state (b) at the final transverse stretching stage (Picture 5-7b) (Pezowicz, 2010).

Pic 5-7: (a) Annulus lamellas in tensile, (b) Annulus lamellas in unloaded position

- Increasing the tensile force causes to decrease the size of the opening in the annulus tissue (Singha & Singha, 2012).

- Intervertebral discs are “shock absorbers”, so they are too stiff in compression (Adams, 2015).

- The compression of spinal is 3–5 kN during manual hand working (Adams, 2015).

- The greatest compressive forces on the spine usually arise when body weight is speeded up (during pilot ejection from an airplane and fall someone on their buttocks). It can increase spinal compression by more than 100 % compared to the same movement in performing slowly (Adams, 2015).

- The external stresses value is zero MPa in an unloaded control level, between the 0.2-0.4 MPa in a sitting pressure (close to body weight exertion), 0.5 MPa for walking, 0.1 MPa for the resting pressure, and 1 MPa as the external stresses (300% of body weight being exerted) (Figure 5-32a) (Gabai, 2007).

- The overload of compressive forces leads to fracture of the endplate of vertebrae before any damage to the disc. So bone mineral density (BMD) has an important role in the fracture in compressive loads (Adams, 2015).

- In the compressive stress (load), the radial and longitudinal pressure on annulus tissue are increased proportionally (Singha & Singha, 2012).
- Simple shear-stress plot for annulus (Figure 5-32b) (Singha & Singha, 2012).

Fig 5-32: (a) Stress- time curve in different types of loadings, (b) Shear- stress curve of annulus

- The curve between displacement and time of loading on annulus tissue: the rapid deformation due to the large amounts of fluid (water) is going out from the annulus tissue, a constant value of displacement and displacement slows down due to the fluid flow slows to zero (Figure 5-33) (Singha & Singha, 2012).

Fig 5-33: Displacement- time curve in compression of annulus

- The total compression stress on the disc made by compression of different parts of the disc (Singha & Singha, 2012) (Formula 5-4):

$$W_{total} = W_{matrix} + W_{shear\ int.} + W_{fibre} + W_{normal\ int.} \quad (5-4)$$

- The shear modulus value is less than the tensile and the compression modulus values because the shear force is a biaxial (X-Y plane velocity) but the compression and tensile is uniaxial.

- Schematic representation of various mechanical testing for goat AF tissue (Figure 5-34):

Fig 5-34: Schematic representation of various mechanical testing for goat AF tissue

-The torsion modulus is much lesser than another compression, tensile and shear modulus values because of the rotating movement of the annulus tissue is in a short angle (14.5°).

5.1.3.4 Factors affecting the mechanical behavior

- **Degeneration** (decreasing hydration of the disc)

Degeneration is happened by genetics, mechanical factors, aging (Gabai, 2007), not enough compression on the disc, the stress of structural damage, and impairs disc cell metabolism. (Table 5-9) (Adams, 2015).

Degeneration causes a decrease in permeability (Gabai, 2007), the ability to repeat, less metabolically active, radial fissures in the annulus, disc herniation, endplate fracture, internal collapse of annulus lamellae (Adams, 2015), the pressure towards the surrounding nerves (Singha & Singha, 2012), biochemical changes, nucleus pressure reduction to to zero and negative pressures in spine traction or in bending backward (Adams, 2015), change of the tensile stresses in the collagen fibers to compressive stresses, and if compressive forces are high and the nucleus pulposus of disc is incompressible the force converted to the annulus (Figure 5-35a) (Jongeneelen et al., 2006).

Table 5-9: Experimental data of strength of lumbar intervertebral disc

Loading	Site of failure	Strength (STD)
MS- compression	Endplate	10,2 (1.7)kN
MS- Shear	Neural arch	2 kN
MS- Forward Bending	Posterior ligaments	73 (18) Nm
MS- Backward bending	Neural arch	26-45 Nm
MS- Torsion	Neural arch	25-88Nm
Disc- compression	(see above for MS)	
Disc- Shear	annulus	0.5 kN
Disc- flexion	Posterior annulus	33 (13) Nm
annulus	37.2+3.2	10-31 Nm

In the normal disc (left), the neural arch resists only 8% of the applied compressive force and distribute the force fairly between the anterior and posterior. But in disc degeneration (right), the neural arch resists 40% of the applied compressive force whereas the anterior vertebral body resists only 19% of that (Singha & Singha, 2012). So degeneration has an influence on the distribution of compressive force between the anterior and posterior of the disc (Figure 5-35b) (Table 5-10) (Adams, 2015).

Fig 5-35: (a) Tensile loading on normal and degenerative disc, (b) distribution of load in normal and degenerative disc

The nucleus pulposus (NP) can bear compressive loads daily so it shows early signs of degeneration (typically 30 %) (Figure 5-36) (Adams, 2015).

Table 5-10: Degeneration distribution of compressive force on the disc

Condition of disc	Distribution (%) of applied compressive force		
	Anterior disc	Posterior disc	Neural arch
Normal/mature	32 (27)	49 (25)	19 (8)
Mild degeneration	21 (20)	41 (20)	38 (18)
Severe degeneration	10 (8)	12 (8)	78 (9)

Fig 5-36: compressive load curve on normal and degenerative disc

- Hydration

Hydration level and the height of the disc have a significant impact on disc stiffness, time-dependent behavior, and mechanical behavior (Gabai, 2007).

- Aging

The density of cell Disc decreases during growth and it gets stable around the age of 20 years.

Aging changes appear first in the disc nucleus such as loss of proteoglycans and fragmentation, reduces the water content of the nucleus, makes hydrostatic pressure and more compressive, increases collagen type I in the inner annulus, makes fibrous texture, makes cartilage stiffer, causes more easily to injury, increases displacements of annulus lamellae, increases tensile strains, and makes disc color brownish (Picture 5-8) (Adams, 2015).

Pic 5-8: Disc by aging

Nucleus becomes stronger, stiffer and less deformable by aging, but its collagen decreasing in degenerative conditions could make an opposite impact (Adams, 2015).

That aging makes annulus weaker in the macroscopic tissue between the annulus lamellae, but make the annuals stiffer and less deformable (Adams, 2015).

Annulus strength decrease with age based on its density (by 43 %, from 40 to 24 Nm). The spine becomes weaker, stiffer and less mobile in parallel of decreasing strength in ligaments and bones (Adams, 2015).

The AF tissues is a perfect composite material because it needs multiple cracks to face its final failure, while the other homogeneous structure fails with a single crack. Annulus tissue is starting to damage by leaving the fibers out of the matrix, matrix declining by cracking, and facing over longitudinal tension, then damaging of fibers buckling accumulation and leading to final failure of the tissue happens (Adams, 2015).

- Herniation

Herniation is happened in these two ways and makes discs to fail:

- When the nucleus has a high water content but the annulus is weakened because of age (40–50 years).
- By less severe and cyclic loading in compression and bending (Adams, 2015).

- Damage

Compressive damage makes a pressure reduction in the nucleus and high-stress peaks in the annulus (Figure 5-37).

Fig 5-37: compressive- distance between disc in compression damages

The mechanical behavior of the disc depends on two basic mechanisms: the mechanical of the annulus fibrosus fibers and the pressure between the outside and inside of the disc by the hydrating effects of nucleus pulposus proteoglycans (Gabai, 2007).

5.1.4 Tendons and ligaments

Tendons and ligaments are generally long and can be tested in their isolated state using sinusoidal-shape or frozen grips to limit slippage (Woo, Almarza, Karaoglu, & Abramowitch,

2008). Ligaments and tendons are soft connective tissues (Reese, 2012) for transferring tensile loads, controlling motion and stabilizing joints in the skeletal system (Woo et al., 2008). Ligaments task is connecting bone to bone and tendons task is connecting muscle to bone (Reese, 2012).

The spinal ligaments task is making the stability of the spinal column, protecting the spinal cord (Hirsch, Ingelmark, & Miller, 1963), bearing high loads, dissipating energy, and interfacing with various materials with different stiffness such as bone and muscle (Reese, 2012). We have faced the joints activities in the musculoskeletal system by sending data of neurological feedback signals of ligaments which are strained to activate the muscles (Frank, 2004). The ventral and dorsal longitudinal ligaments are in spinal bodies. The ventral ligament consists of dense fibrous which are connected to the annulus fibrous, but the dorsal ligament is thinner without attaching to the disc (Hirsch et al., 1963).

The ligaments of the vertebral arches include the supraspinous group, the intertransverse, and the ligament flavum.

The supraspinous group is a true ligament (Hirsch et al., 1963). The intertransverse ligaments appear more as a part of the lumbar-dorsal fascia than as true ligaments (Hirsch et al., 1963). The flavum ligament has the structure and function of a ligament but it also acts as a fibrous capsule on the ventral surface of the lumbar synovial joint and as an elastic band keeping the spinal nerves free from compression when passing through the inner vertebral foramen during movements in the lumbar spine (Hirsch et al., 1963).

5.1.4.1 Ligaments types

The most ligaments of the spine consist of:

- The anterior longitudinal ligament (ALL) stabilizes the spinal column by connecting adjacent anterior aspects of the vertebral bodies.
- The posterior longitudinal ligament (PLL) stabilizes the spinal column by connecting adjacent posterior aspects of the vertebral bodies.
- The ligament flavum (LF) prevents potentially damaging hyperflexion motion by connecting adjacent lamina (Troyer & Puttlitz, 2011).

The longitudinal ligaments (ALL and PLL) have similar morphologies that consist of a large amount of highly oriented collagen fibers around 1:11 elastin-to-collagen ratio. Conversely, the LF has a 2:1 elastin-to-collagen ratio (Troyer & Puttlitz, 2011). The longitudinal ligaments (ALL and PLL) are narrow fibrous bands with viscoelastic behavior to stabilize the spinal column during flexion and extension motions and to maintain intradiscal pressures by resisting disc bulge (Troyer & Puttlitz, 2012). The LF is a thick and wide structure with high elastin fibrous content and less viscous behavior than the longitudinal ligaments. So LF has more increased elastic behavior than the others and it has no mechanical functions too much (Troyer & Puttlitz, 2012). The ALL has the most viscoelastic behavior than the other longitudinal ligaments. Therefore, it has more motion, rotation, and stress than the PLL. For example, in extension, ALL experiences a greater amount of deformation and stress than the

PLL inflection (Troyer & Puttlitz, 2012). The posterior longitudinal ligament is usually the weakest (Myklebust, Sances Jr, Maiman, Pintar, & Chilbert, 1982).

5.1.4.2 Ligaments and tendons materials

Tendons and ligaments are packed with soft connective tissues of parallel collagen fiber bundles (“Introduction to Biosolid Mechanics,” n.d.; Woo et al., 2008) for transmitting loads within the musculoskeletal system (Reese, 2012).

They are made of a hierarchical structure. The ligament or tendon (at the macroscale) is contained of fascicles in parallel to span the entire bone-bone for ligament or muscle-bone for tendon length (at the mesoscale) (Reese, 2012). Each fascicle has the basic fibrils (biological cells (at the nanoscale) of the ligament or tendon (“Introduction to Biosolid Mechanics,” n.d.), and fibrils form fibers (at the microscale) (Figure 5-38) (Reese, 2012).

Fig 5-38: Ligaments and Tendons components

The basic fibrils have structural character and significant role in the mechanics of ligaments and tendons because of its crimp period (range from 50 to 200 μm) and angle (Reese, 2012). The crimp causes the waviness of the fibril, incredible strength and makes the nonlinear stress-strain relationship for ligaments (“Introduction to Biosolid Mechanics,” n.d.).

Tendons and ligaments undergo a fluid phase and a solid phase (the bulk of the tissue consisting of 65%-75% water by weight) (Reese, 2012), and they have hierarchical structure of collagen type I (70-80% of the dry weight), proteoglycans (<1%), elastin, glycolipids, water (65e 70% of the total weight), cells (Woo et al., 2008), and other proteins and glycoproteins such as actin, laminin, the integrins (Frank, 2004), and other collagen types (III, V, X, XI, and XII) (Woo et al., 2008).

The water and proteoglycans provide lubrication and space for gliding function of fibers. Elastin allows the tissue to return to its pre-stretched length following the physiological loading. Collagen type V exists as a regulator of collagen type I, collagen type III is for wound

healing, collagen type XII provides lubrication between collagen fibers, and collagen types IX, X, XI, and II collagen at the fibrocartilaginous zone for minimizing the stress when loads are transmitting from soft tissue to bone (Woo et al., 2008). Proteoglycans (with small quantities of (0.5%-3%)) play an important role in tendon and ligament mechanics. It contains decorin (~1%/wt), biglycan (~0.5%/wt) and others such as fibromodulin, lumican, aggrecan, and versican. The absence of decorin (consists of a core protein) leads to altered fibrillar organization and a decrease in mechanical properties such as strength and stiffness. Biglycan plays a similar role in facilitating the organization of collagen, but its role is less clear. Aggrecan and versican are present in small quantities and may play roles in maintaining tissue hydration via osmotic swelling (Reese, 2012). Elastin occurs at 1-2% by dry weight, is found in the form of elastic fibers. In mature tissue, elastic fibers (d~2 μ m) consist of a core of amorphous elastin covered in microfibrils which are 10-12 nm fibrils composed of fibrillin (Reese, 2012).

Ligaments and tendons are morphologically similar to each other, but there are some biochemical differences. Ligaments are more metabolically active, having more cellular nuclei, higher DNA content and greater amounts of reducible cross-links between collagen fibers. Tendons, on the other hand, contain less water, less a proteoglycan matrix, and slightly more type I collagen with much smaller amounts of other collagens, such as collagen type III, V, XII, and XIV (Woo et al., 2008). Therefore, their differences are in fibril diameter distributions, crimp morphologies, metabolic activity, and the percentage of components such as water, proteoglycans, and types I and III collagen. Structure and composition do not only vary between tendons and ligaments but also vary between ligament and tendon locations (Reese, 2012).

5.1.4.3 Ligaments and tendons biomechanics

- The motion capacity of the vertebral column varies along its length. The mobility of the cervical vertebrae is much higher than that of the thoracic or lumbar ones. The upper cervical spines are more flexible than the lower cervical levels (Polak, Czyż, Ścigała, Jarmundowicz, & Będziński, 2014).
- Stabilizing joints passively through their normal range of motion in a tensile loading is the function of ligaments. Ligaments have this nonlinear anisotropic viscoelastic mechanical behavior due to the crimp of its collagen fibers (Frank, 2004).
- Guiding the joint motions by transmitting forces along their longitudinal axis to make anatomical skeleton alignment, the balance of the direction and magnitude of externally applied loads, and contact between joint surfaces, ligament forces, and muscle activity (Woo et al., 2008) are the roles of ligaments.
- The hierarchical structure of ligaments and tendons causes both nonlinear and viscoelastic behavior of ligaments, make them more difficult to analyze and quantify, and has a significant influence on mechanical behavior ("Introduction to Biosolid Mechanics," n.d.).
- Ligaments and tendons have time and history-dependent viscoelastic behavior due to complex interactions of tissue constituents such as collagen, proteoglycans, water, and ground

substance (Woo et al., 2008). The stability of the spinal column, ability to protect the spinal cord from excessive motion and extra energy, facilitate three-dimensional physiologic motion patterns and maintain vertebral postures during both static and dynamic loading regimes are all because of its time-dependent (viscoelastic) mechanical behavior of the spinal ligaments (Troyer & Puttlitz, 2011, 2012). This phenomenon reduces the capacity of damage which is related to its fatigue (Woo et al., 2008).

- Viscoelasticity indicates the time and history-dependent mechanical behavior. Thus, the relationship between stress and strain is not constant, depends on the time of displacement or load, has higher strain rates result in stiffer behavior, and has slower strain rates result in more compliant behavior (Reese, 2012).

- Viscoelasticity makes these three behavior characteristics:

- Creep has an increasing deformation under constant load in contrast with an elastic material which does not exhibit increase deformation depend on time (Figure 5-39a) (“Introduction to Biosolid Mechanics,” n.d.).

- The stress will be reduced or will relax under a constant deformation (stress relaxation) (Figure 5-39b) (“Introduction to Biosolid Mechanics,” n.d.).

Fig 5-39: (a) Creep curve of ligaments and tendons, (b) stress relaxation

- If a viscoelastic material is loaded and unloaded, the unloading curve will not follow the loading curve due to the loss of energy during loading (hysteresis or energy dissipation). This causes of pseudo-elasticity and the nonlinearity (Quasi-linear Viscoelastic Theory) (Figure 5-40a) (“Introduction to Biosolid Mechanics,” n.d.; Woo et al., 2008).
- The spine ligaments exhibit nonlinear. So, their stress behavior is dependent on their strain magnitude and each ligament type (Figure 5-40b) (Troyer & Puttlitz, 2012).
- A typical stress-strain curve for ligaments and tendons has three major parts (if viscoelastic behavior neglects):
 - The toe or toe-in region (linear section represents collagen fibers in relaxed condition).
 - The linear region (all the fibers are stretched and the curve has a linear shape).

Fig 5-40: (a) Stress- strain curve in loading, unloading and reloading, (b) stress- strain curve of different ligaments

- The yield and failure region (Overload and the ultimate tensile strength is reached) (Figure 5-41a) (“Introduction to Biosolid Mechanics,” n.d.; Polak et al., 2014).

In the physiologic activity, most ligaments and tendons exist in the toe and somewhat in the linear region. These constitute a nonlinear stress-strain curve since the slope of the toe-in region is different from the linear region (“Introduction to Biosolid Mechanics,” n.d.).

- The crimp structure and failure of the collagen fibrils make the ligaments and tendons behavior. In the toe-in region because of un-crimping of the collagen fibrils, the stress-strain curve shows a low stiffness. After that, a ligament or tendon is stretched to its limit so its stiffness increases until the ligament or tendon begin to fail. As a result, in fail damages step its stiffness is reduced again (“Introduction to Biosolid Mechanics,” n.d.).
- The elastic behavior of ligament is anisotropic and nonlinear.
- The ligaments are denticulate to stabilize the spinal cord with intervertebral canal and prevent overloading of spinal nerve roots during their excessive overstretching due to the connection between the side surface of the spinal cord and the spinal canal wall (Polak et al., 2014).
- The averaged stress-strain curve of ligaments from different cervical levels (Figure 5-41b) (Polak et al., 2014).

Fig 5-41: (a) The main Stress- strain curve, (b) Stress- strain curve of different cervical ligaments

- The average ultimate force that breaks the denticulate ligaments was 0.88N (with a decrease of the failure force between the C1 and C7 (With interrupted at the level of the C4 vertebra) (Polak et al., 2014).
- The lowest maximum stress and failure force values for the denticulate ligaments were obtained for the lowest level of the cervical spine and was 1.31MPa (Polak et al., 2014).
- The ultimate forces for the single vertebral bodies in compression ranged from 1500-7400N (Polak et al., 2014).
- The mean Young's modulus was 2.06MPa with a minimum of 1.31MPa for C7 and maximum of 2.46MPa for C5 (Polak et al., 2014).
- The highest elastic modulus was obtained for ligaments at the C5 level (2.46MPa) as the only deviation from the distribution (Polak et al., 2014).
- The mean toe forces for all of the samples were 0.47N. Before reaching the breaking force, samples were extended to nearly twice lengths of its initial size (2.0mm) (Table 5-11) (Polak et al., 2014).

Table 5-11: Average Toe and Failure results for different cervical spine parts

Level	Toe Elongation (mm)	Toe force (N)	Toe strain	Toe Stress (MPa)	Failure elongation (mm)	Failure force (N)	Failure strain	Failure stress (MPa)
C1	0.97 (0.60)	0.30 (0.19)	0.49 (0.29)	0.67 (0.40)	2.53 (1.07)	1.04 (0.41)	1.27 (0.51)	2.31 (0.88)
C2	1.00 (0.57)	0.26 (0.19)	0.50 (0.27)	0.58 (0.33)	2.56 (1.35)	1.04 (0.68)	1.28 (0.65)	2.31 (1.45)
C3	0.88 (0.44)	0.17 (0.08)	0.44 (0.21)	0.37 (0.18)	2.49 (0.87)	0.88 (0.25)	1.24 (0.42)	1.95 (0.54)
C4	0.87 (0.34)	0.31 (0.25)	0.43 (0.16)	0.69 (0.52)	2.19 (0.77)	1.04 (0.50)	1.10 (0.36)	2.31 (1.03)
C5	0.60 (0.25)	0.18 (0.08)	0.30 (0.12)	0.40 (0.18)	2.04 (0.78)	0.84 (0.37)	1.02 (0.37)	1.87 (0.77)
C6	0.73 (0.28)	0.17 (0.11)	0.36 (1.13)	0.37 (0.23)	2.29 (0.81)	0.74 (0.20)	1.14 (0.38)	1.65 (0.43)
C7	0.70 (0.31)	0.09 (0.05)	0.35 (0.14)	0.21 (0.09)	2.34 (1.16)	0.55 (0.12)	1.17 (0.54)	1.22 (0.25)
Average	0.82 (0.14)	0.21 (0.07)	0.41 (0.19)	0.47 (0.28)	2.35 (0.22)	0.88 (0.19)	1.17 (0.46)	1.95 (0.76)

- The stiffness is ranged from 3550 to 19700 N/cm with a mean of 9520 N/cm (Myklebust et al., 1982).
- The stiffness in tensile along the fiber direction is a stiffer than in the transverse direction of magnitude (left), in shear is two times more compliant than transverse direction (center) and in compression is over three times more compliant than in tension (right) (Figure 5-42) (Reese, 2012).
- The average strength of the anterior longitudinal ligament (ALL) ranges from 180 to 500 N, with the strongest values in lumbar levels.

- The average strength of the posterior longitudinal ligament (PLL) ranges from 45 to 160 N. with the strongest at mid thoracic levels.

Fig 5-42: Compare of tensile, compression, and shear stress of ligaments

- The average strength of the ligament flavum (FL) ranges from 90 to 270 N. with the strongest at mid lumbar levels.
- The average strength of the interspinous ligament demonstrated a range of 45-200 N with the highest values in the mid-thoracic spine (Myklebust et al., 1982).
- The range of force for disruption of single ligaments (Table 5-12).

Table 5-12: The range of force of ligaments

Ligaments	The range of force
Anterior longitudinal	185- 500 N
Posterior longitudinal	45- 160 N
Joint capsules	90- 380 N
Ligaments flavum	50- 270 N
Interspinous Ligaments	45- 200 N

- Increasing flexion causes increasing loads of ligaments up to the point of failure (Myklebust et al., 1982).

- The material properties of ligaments (Table 5-13) (Kumaresan, Yoganandan, Pintar, & Maiman, 1999; Toosizadeh & Haghpanahi, 2011).

Table 5-13: Nonlinear material properties of ligaments

ALL Anterior Longitudinal Ligament		PLL Posterior Longitudinal Ligament		ISL Interspinous Ligament		LF Ligamentum Flavum		CL Capsular Ligament	
Def. (mm)	Force (N)	Def. (mm)	Force (N)	Def. (mm)	Force (N)	Def. (mm)	Force (N)	Def. (mm)	Force (N)
1.4	35.5	1.0	29.0	1.3	16.9	1.9	45.9	1.8	53.6
2.7	64.9	2.0	51.4	2.7	24.4	3.7	82.4	3.9	87.9
4.1	89.7	3.0	71.3	4.0	29.5	5.6	119.6	5.8	109.4
5.4	108.6	4.0	85.5	5.4	32.9	7.5	133.7	7.7	125.8
6.8	119.6	5.0	94.7	6.7	34.9	9.4	147.2	9.7	134.8

5.1.4.4 Factors affecting the mechanical behavior

- **Injury:** ligaments and tendons usually are torn partially or completely in joint injuries (Frank, 2004) and their injuries remain difficult to treat and more slowly to change (“Introduction to Biosolid Mechanics,” n.d.) due to their poor vascular and cellular. After scar tissue is formed, it is structurally and mechanically different from normal tissue (Figure 5-43a) (Reese, 2012).

Treating ligaments happens in two ways:

- Immobilization of a joint for a long period of time leads to significant changes in joint structure and function, including decreased range of motion for the joint, reduction in mechanical properties, and reduction in the ligament structure. Upon remobilization, it appears that the mechanical properties of the ligament are gained back first followed by the structural properties (Figure 5-43b) (“Introduction to Biosolid Mechanics,” n.d.).

Fig 5-43: (a) Stress- strain of ligaments by injury, (b) Treating by immobilization and exercise for ligaments

- Exercise and increased load on tendons and ligaments increase mechanical properties (22% increase in tensile strength) (“Introduction to Biosolid Mechanics,” n.d.). It has a less rapid and substantial effect on mechanical properties than immobilization.

- **Damage:** tendon and ligaments damage and failure appears and occurs at the fascicle and fiber level (Reese, 2012).

- **Aging:** increasing age from child to adult causes to increase the mechanical properties of ligaments and tendons, but increasing age from young adulthood causes to decrease the properties of ligaments and tendons (2-3 times) (“Introduction to Biosolid Mechanics,” n.d.).
- **Healing:** the healing of tendons and ligaments happens by increasing amounts of proteoglycans, the higher ratio of type V to type I collagen, and decreasing in the number of mature collagen cross-links (Type III collagen is a type of collagen associated with wound healing) (Woo et al., 2008).

5.1.5 Facet Joints

The facet joints, the intervertebral discs, and spinal ligaments are connecting the adjacent vertebrae of the spine in all regions to support spine for the transfer, constraint of loads applied to the spinal column, ensure the mechanical stability, overall mobility of the spine, and protect the spinal cord running through it.

5.1.5.1 Facet joints components

The zygapophyseal, the facet joint, is such a synovial joint that has a crucial biomechanical role in the spinal column because of having a variety of hard and soft tissues. It causes to resist traumatic and physiologic loads, maintain strength, make the spine stable (Kumaresan, Yoganandan, & Pintar, 1998), and guide and steady the movements of each segment (Wu & Chen, 1996).

The cervical spine facet joint consists of the facet bone, capsular ligament, synovial fluid, synovial membrane, and articular cartilage. Facet bones are in the opposing surfaces (diarthrodial joint) with articular cartilage surfaces for providing a low friction environment, the synovium (a connective tissue lining) for maintaining lubrication, and a ligamentous capsule for enclosing the joint space (Jaumard, Welch, & Winkelstein, 2011).

5.1.5.1.1 Facet bones

Facet bones are part of a vertebra, which has both an inferior and a superior elliptically shaped surface in its articulate with cartilage on both of them. The morphology and the angles of these surfaces depend on their level of vertebrae. The angles of the facet bones in the sagittal plane are ranged from 20-78, 55-80, and 82-86 in the cervical, thoracic, and lumbar region respectively (Figure 5-44). The angles between them in the axial plane are ranged from 70-

Fig 5-44: vertebrae structure

96, 85-120, and 15-70 in the cervical, thoracic, and lumbar regions respectively (Figure 5-45a) (Jaumard et al., 2011).

The superior and inferior facet of the vertebra is flat in the cervical and thoracic regions but the superior is more convex and the inferior is more concave in the lumbar region (Jaumard et al., 2011).

5.1.5.1.2 Cartilaginous

Cartilaginous is an avascular layer of transparent cartilage which is covered the surfaces of each facet bones with different thickness according to its spinal regions and genders (not uniform) (Jaumard et al., 2011). The cartilage is thinner at the edges and gradually its thickness increases (1 mm) towards the center of the surfaces where both the maximum thickness and the shape coefficient are (ranging 0.38–0.63) (Jaumard et al., 2011). The articulating facet surfaces make a frictionless motion in facet joints, also loading compressive, tensile, and shear.

The cartilage matrix consists of collagen fibers, glycosaminoglycan's (GAGs), chondrocytes, and proteoglycans. Each cartilage zone has different structural parts such as (Figure 5-45b):

Fig 5-45: (a) Facet joints in axial plane, (b) Cartilaginous matrix

- The superficial zone: it contains few oriented tangential horizontal flattened chondrocytes and collagen fibers at the surface of the cartilage to provide resistance according to the tensile and shear stresses during bending or rotating.
- Transitional zone (middle zone): there are more chondrocytes with an accidental arrangement between larger collagen fibers.
- The deep zone: chondrocytes are aligned in columns perpendicular to the articular surface with the parallel of the collagen fibers.

The transitional and deep zones because of having the concentration of proteoglycans are trapping water and increasing the incompressibility of the structure, so they are supporting compressive and hydrostatic stresses.

- The tidemark zone: the calcified cartilage and subchondral bone at the bottom of the deep zone which is changed from the elastic modulus of cartilage to bone gradually (Jaumard et al., 2011).

The mechanical properties of cartilages are heavily dependent on its composition (water content, collagen fiber orientations).

The balanced modulus (H) of cartilage in compression is decreased linearly with age, structural disorganization (I), and importantly water content (WAT) (a 30–40% decrease in water content leads to a 161% increase in the balanced elastic modulus of cartilage) (Formula 5-5).

$$H = 1.04 - 0.0045 \cdot (\text{Age}) \quad (5-5)$$

$$H = 0.95 - 0.065 \cdot (I)$$

$$H = 5.29 - 0.058 \cdot (\text{WAT})$$

Articular cartilage is a composite material of fluid (water) and solid (chondrocytes, collagen, and proteoglycans) phases. Therefore, it has anisotropic nonlinear biphasic and time-dependent mechanical properties with loadbearing capacity (Jaumard et al., 2011).

5.1.5.1.3 Synovium, menisci, and capsular ligament

- Between two opposing articular pillar, there are the **synovium** and the ligamentous capsule to maintain a low-friction environment and provide mechanical resistance to their separation and motion. The synovium is a thin and soft connective tissue to secrete synovial fluid to lubricate and nourish the cartilaginous articular surfaces. The synovium is difficult to isolate and the internal synovial layer is easy to lose so their mechanical behavior is difficult to find (Jaumard et al., 2011).

- **The synovial folds (meniscoids or menisci)** are composed of fat and fibrous connective tissue. They are intra-articular structures for protecting the articular cartilage in gliding by opposite articular surface and covering the gap of exposed subchondral bone to reduce friction during articular motion. Menisci have three main types across all of the regions of the spine: adipose tissue pads, fibro-adipose meniscoids, and connective tissue rims (Jaumard et al., 2011).

- **The facet capsular ligament** covers the synovium to enclose the facet joint in the superior-inferior direction with a thickness of 2.0 mm in the posterior region and 3.2 mm in the anterior region. The capsular ligament is contained dense collagen fiber, which is bundled with proteoglycans, elastin fibers, and fibroblasts interspersed. The collagen and elastin fibers extend between the laminae of adjacent vertebrae connecting to the ligament flavum both in the interior and posterior regions of the facet joint and completely surrounding the joint's articular surfaces in three dimensions. The crimped collagen fibers allow the capsule to work without injury and under loading allow joints to rotate by being crimped to resist mechanically (Jaumard et al., 2011).

5.1.5.2 Facet joint mechanical behavior

Fig 5-46: The maximum and average stress and strain of the facet joint (L4-L5) for normal and degenerative lumbar

- The maximum and average stress and strain values of articular cartilage on the facet joint of L4-L5 for the normal lumbar and degenerative lumbar (Figure 5-46) (Du et al., 2016).
- The maximum facet joint stress in extension (Figure 5-47a) (Li et al., 2017).
- Facet joint stress distribution at cartilage between C5-C6 in extension (Figure 5-47b) (Li et al., 2017).

Fig 5-47: (a) The maximum facet joint stress in extension, (b) Facet joint stress distribution at cartilage between C5-C6 in extension.

- The quantity of load which is carried by the facet joint depends on the inter-facet joints gap and the level and number of motion segments (Kumaresan et al., 1998).

- The facet capsular ligament is composed of elastin and collagen fibers, so it can support tensile and shear loading, limit the motions of the facet joint, spinal stability, guide and constrain the motion of the vertebrae, prevent the intervertebral disc and the nerve roots from damage, provide torsional stiffness, resistance to shear, lateral, and anterior-posterior vertebral translation, joint distraction, and facilitate the transmission of the loads applied to the spine (3–25% of the compressive load to the lumbar spine). The spine becomes unstable when translations are greater than 3.5 mm and rotations greater than 20 degrees in the sagittal plane during flexion-extension bending (Jaumard et al., 2011).

- Average failure strains are ranged from 100 to 150% for the cervical capsular ligament.

- The strain and the stress of the viscous (V) and elastic (E) cases (Jaumard et al., 2011) (Formula 5-6):

$$\text{Parallel (viscos), } \sigma_Y = 0.0034 \cdot e^{(10.35 \cdot \epsilon)} \quad (5-6)$$

$$\text{Perpendicular (viscos), } \sigma_Y = 2.02 \cdot \epsilon - 0.1732$$

$$\text{Parallel (Elastic), } \sigma_Y = 0.0030 \cdot e^{(10.09 \cdot \epsilon)}$$

$$\text{Perpendicular (Elastic), } \sigma_Y = 1.04 \cdot \epsilon - 0.097$$

- The facet joints have a nonlinear relationship in its cervical and lumbar spines part (maximum principal strains were 12% in flexion and extension) (Jaumard et al., 2011).

- The facet capsule mechanical properties in the cervical and lumbar spines part (Table 5-14) (Jaumard et al., 2011).

Table 5-14: The facet capsule mechanical properties in cervical and lumbar spines part

level	Loading details		Property	
	Direction	magnitude		
C2- C5	Axcial Tension	Failure	Failure strain (%)	148(+_- 28.5)
			Failure stress (MPa)	5.67 (+_- 1.47)
		Failure	BILinear muduli (MPa)	E1= 5.0, E2= 3.3
			Transition strain	€12= 56.8
			Stiffness (N/mm)	33.6 (+_- 5.53)
C5- T1	Axcial Tension	Failure	Failure strain (%)	116(+_- 19.6)
			Failure stress (MPa)	7.36 (+_- 1.27)
			BILinear muduli (MPa)	E1= 4.8, E2= 3.4
			Transition strain	€12= 57.0
			Stiffness (N/mm)	36.9 (+_- 6.06)
C3- C4 C5- C6	Shear + Axcial compression	0N	Maximum principal strain (%)	17.0 (+_- 6.4)
		45N		15.5 (+_- 5.8)
		197N		16.8 (+_- 6.1)
		325N		17.8 (+_- 5.8)
	Shear	Failure	Subcatastrophic failure strain (%)	35 (+_- 21)
			Catastrophic failure strain (%)	94 (+_- 85)
C3- C4 C5- C6	Flexion	2.5 Nm (flex) +	Strain (%)	12.1 (+_- 2.5)
				23.0 (+_- 4.4)

	Flexion + contra axial pretorque	1 Nm (torque)		15.6 (+_ 6.8)	
	Flexion +ipsi axial pretorque				
	Extension	2.5 Nm (Ext)		11.6 (+_ 2.6)	
	Extension + contra axial pretorque	+		17.2 (+_ 5.8)	
	Extension +ipsi axial pretorque	1 Nm (Torque)		12.6 (+_ 4.2)	
		Failure	Subcatastrophic failure strain (%)	64.6 (+_ 73.8)	
			Catastrophic failure strain (%)	103.6 (+_ 80.9)	
L3- L4 L4- L5	Flexion	15 N	Strain (%) in active zone (i.e. outside of the neutral zone)	Left	Right
				13.0(+_5.5)	10.4(+_5.6)
	Extension	15 N		3.7 (+_1.7)	6.0 (+_ 2.1)
	Left torsion	15 N		8.7 (+_ 4.3)	-3.6 (+_ 2.7)
	Right torsion	15 N		-3.3 (+_ 2.7)	7.0 (+_ 1.8)
	Left lateral bending	15 N		8.8 (+_ 5.4)	1.8 (+_ 4.7)
T12- S1	Right lateral bending	15 N		3.2 (+_ 4.0)	1.2 (+_ 4.3)
	Flexion	3 Nm	Maximum principal strain (%) At 40 mm of horizontal displacement of T12	14.6	
	Extension	5Nm		7.7	
	Right lateral bending (4.5Nm)	4.5 Nm		9.4	
	Left lateral bending (5.5 Nm)	5.5 Nm		8.6	
Li- S1	Axial tension	50% strain	Modulus(MPa) Tangent	8.08 (+_ 5.75)	2.76 (+_ 1.32)
			Modulus(MPa) viscos	1.61 (+_ 0.36)	1.81 (+_ 0.29)
			Modulus(MPa) elastic	1.33 (+_ 0.49)	1.00 (+_ 0.27)
			Poisson ratio	0.299(+_ 0.15)	0.488 (+_ 0.199)
C6- C7	Subfailure distraction axial		Stiffness(N/mm)	1.62 (+_ 0.80)	
		Physiologic (300)	Maximum principal starin (%)	22 (+_ 3)	
		Subcatastroph ic (700)		50 (+_ 12)	
C6- C7	Axial tension	Failure	Failure strain (%)	72.9 (+_ 7.1)	

- The T12-S1 lumbar maximum principal strains are smaller than the cervical spine (15% in flexion, 8% in extension, and 9% in lateral bending).

- The facet capsule can lengthen during shear loading and it is so strong in deformation, rotation, translation, flexion, lateral bending, and axial rotation but it does not stretch in extension in its direction of loading (Jaumard et al., 2011).

- Joint cartilage has viscoelastic properties due to its fluid and solid elements. Collagen fibers provide tensile strength and the water content provides its compressibility (Jaumard et al., 2011).

- The surface of the human facet articular cartilage is curved and has a non-uniform thickness (with a maximum thickness of only approximately 1 mm) (Jaumard et al., 2011).

- The capsular ligament exhibits anisotropic viscoelastic properties.

- Capsular strains are not uniform and vary with the joint loading types (94685% in shear, 104681% to 148628% in tension). Structural damage in the capsule happens at strains of 51612%.
- The Lumbar stresses in the facet capsule are double at 5 degree of flexion.
- The mechanical loading of the facet joint happens by stretching the capsular ligament, compression of the cartilaginous articular surfaces and the subchondral bone (Jaumard et al., 2011).
- The forces on the facet joint are non-uniform depend on spinal regions and the loading types (the facets joints carry 3–25% of the spinal load of axial compression).
- The table of facet forces and pressures on each vertebra (Table 5-15):

Table 5-15: The table of facet forces and pressures on each vertebra

Level	Direction	Magnitude	property				
C4- C5 C5- C6	compression	80N	Force (N)	30-38			
			Pressure (MPa)	0.13- 0.19			
C4- C5	flexion	1.3 (+_0.3) Nm	Pressure (MPa)	0.086 (+_0.012)			
C6- C7	Extension	1.7 (+_0.5)Nm		0.092 (+_0.014)			
	flexion	1.3 (+_0.3) Nm		- 0.047 (+_0.057)			
	Extension	1.7 (+_0.5)Nm		0.158 (+_0.040)			
C5- C6	flexion	2.7 (+_0.3) NM	Pressure (MPa)	0.010 (+_0.010)			
	Extension	2.4 (+_0.3) NM		0.068 (+_0.027)			
C5- C6	compression	73.6 N	Force (N)	4.2N			
	Flexion	1.8 Nm		None			
	Extension			37.6			
	Axial torsion			28.5			
Cervical	compression	4.5 mm	Compressive stress (MPa)	SL	CS	HE	FL
				0.29	0.55	0.39	0.05
	flexion	18 deg		0.04	0.23	3.98	0.28
	extension			0.29	0.30	4.90	0.23
	Lateral Bending		0.02	0.14	3.81	0.23	
T12- L2	flexsion	Up to 7.5 Nm (1.5 Nm increments)	Force (N)	46.1 (+_41.3)			
	extension			51.5 (+_39.0)			
	Axial torsion - ipsi			31.3 (+_33.4)			
	Axial torsion - contra			70.3 (+_43.2)			
	Lateral bending - ipsi			32.0 (+_44.4)			
	Lateral bending - contra			30.6 (+_29.1)			
	Axial compression	400N		45.5 (+_40.4)			
	flexion	Unspecified		46.6 (+_41.9)			
	extension	Unspecified		75.4 (+_39.0)			

L1- L2 L2- L3	Normal position		Pressure (MPa)	4.5 (+_1.6)	
L3-L4 L4-L5	flexion	4 deg		3.7 (+_1.3)	
L5-S1	Extension	4 deg		5.8 (+_1.6)	
		6 deg	6.1 (+_1.9)		
L1- L5	compressive	500 N	Force (N)	43 (@ L2- L3)	
	extension	7.5 Nm (@L1)		86 (@ L2- L3)	
	extension	20 deg (@L1)		117 (@ L4- L5)	
L2- L3	Axial compression	100 N	Force (N)	23 (+_16)	
	flexion	1 N		None	
	Extension	2 N		74 (+_23)	
	Axial torsion	4 N		92 (+_27)	
	Lateral bending - ipsi	1 N		40 (+_32)	
	Lateral bending - contra	1 N		54 (+_19)	
	Axial torsion	2.3 N		32	
		6.0 N		210	
L2- L5	flexsion	7.5 Nm	Force (N)	left	Right
	Extension			2 (+_5)	4 (+_4)
	Lateral Bending			13 (+_14)	14 (+_10)
	Axial torsion			11 (+_11)	16 (+_14)
L4- L5	flexsion	7.5 Nm	Force (N)	None	
	Extension			50	
	Lateral Bending			36	
	Axial torsion			105	
L5- S1	Axial compression	650N			
	Shear	550N			
	flexion	6 deg		Group 1	Group 2
	Extension			40 (+_13)	45 (+_10)
	Lateral bending- ipsi			54 (+_18)	65 (+_18)
	Lateral bending- contra			50 (+_13)	54 (+_19)
		9 (+_4)	33 (+_10)		

5.1.5.3 Factors affecting the mechanical behavior

- **Degeneration:** degeneration is a progressive and gradually condition of modifying the material, biochemical, and structural properties of tissues by aging, injury, infection or inflammation and causes to change the material and mechanical properties of the joint (bony pillars, capsular ligament, primarily on synovium, and cartilage), lead to damage to the materials of the tissues, and affect the mechanical behavior of the whole vertebral segment (Jaumard et al., 2011). Degenerating in cartilage is happened by increasing its water content and decreasing in the cartilage balanced modulus by increasing the solid materials (especially proteoglycans) to resist the compression (Jaumard et al., 2011).

- **Injury:** facet joint injury, spinal fusion, catastrophic stretching, and non-physiologic loading increases spinal mobility and decreases its mechanical properties. Fractures of the bones of

the facet joint leading to joint separation, comminution, split, and traumatic spondylosis, so it requires surgical treatment to reduce its deformity (Jaumard et al., 2011).

5.2 Material properties of the human spine

These types of materials are based on mechanical behavior of human spine but they are useful for making and modeling numerical model of human spine structure in virtual space.

5.2.1 The material properties of each section of the spine

(Table 5-16, 5-17, 5-18, 5-19, 5-20, 5-21)

Table 5-16: The Material properties of C3- C7 (Kumaresan et al., 1998)

Cervical structure	Young Modulus (MPa)	Poisson Ratio
Cortical bone	12000	0.29
Cancellous bone	450	0.29
Endplate	5000	0.40
Posterior element	3500	0.29
Annulus fibers	3.4	0.40
Nucleus pulposus	1	0.49
ALL	30	0.30
PLL	20	0.30
ISL, LF (C1-C2)	10	0.30
SSL, ISL, LF (C2-C7)	1.5	0.30
CL (C1-C3)	10	0.30
CL (C3- C7)	20	0.30
CL (C0-C1)	1	0.30
AIL	5	0.30
TL	20	0.30
Apl	20	0.30
Anterior membrane	20	0.30
Posterior membrane	20	0.30
NL	20	0.30

Table 5-17: Material Properties for Various components (Li et al., 2017)

Cervical structure	Young Modulus (MPa)	Poisson Ratio	Cross Section Area (mm ²)
Cortical bone	12000	0.29	-
Cancellous bone	100	0.29	-
Cartilage	10.4	0.4	-
Annulus ground substance	3.4	0.4	-
Annulus fibers	450	0.45	-
Nucleus pulposus	1	0.49	-
Endplate	1200	0.29	-

Anterior longitudinal	10	0.3	12
Posterior Longitudinal	10	0.3	45
Capsular	10	0.3	14
Flaval	1.5	0.3	5
Interspinous	1.5	0.3	13

Table 5-18: Material properties of the spine (Zhang, Teo, Ng, & Lee, 2006)

Cervical structure	Young Modulus (MPa)	Poisson Ratio
Articular pillar	3500	0.25
Cartilage	10.4	0.4
Hyperelastic solid	2000	-
Hyperelastic fluid	1000	-
Synovial membrane	10	0.40
Capsular ligament	Deflection (mm)	Force (N)
	0	0.30
	1.7	1.5
	3.6	29.1
	5	52.4
	7.5	85.8
	9.5	104
	9.9	107.9

Table 5-19: Material properties of the spine (Schmidt et al., 2006; Toosizadeh & Haghpanahi, 2011)

Cervical structure	Young Modulus (MPa)	Poisson Ratio
Cortical bone	$E_{xx} = 11300$	$V_{xy} = 0.484$
	$E_{yy} = 11300$	$V_{xy} = 0.203$
	$E_{zz} = 22000$	$V_{xy} = 0.203$
	$G_{xy} = 3800$	
	$G_{xy} = 5400$	
	$G_{xy} = 5400$	
Cancellous bone	$E_{xx} = 140$	$V_{xy} = 0.45$
	$E_{yy} = 140$	$V_{xy} = 0.315$
	$E_{zz} = 200$	$V_{xy} = 0.315$
	$G_{xy} = 48.3$	
	$G_{xy} = 48.3$	
	$G_{xy} = 48.3$	
Bony endplate	$E = 12000-4000$	$V = 0.3$
Cartilaginous endplate	$E = 23.8$	$V = 0.4$
Annulus ground substance	$C1 = 0.56, C2 = 0.14$ ($E = 4.2$ MPa)	$V = 0.45$
Nucleus pulposus	$C1 = 0.12, C2 = 0.09$	$V = 0.4999$
Posterior element	$E = 3500$	$V = 0.3$
Atlas and skull	$E = 2000$	$V = 0.3$
Disc fibers	$E = 5$ (anterior), $E = 1$ (posterior), $E = 0.1$ (lateral)	$V = 0.3$
Alar ligaments	$E = 5$	$V = 0.3$

Transverse Ligaments	E= 20	V=0.3
Apical ligaments	E= 20	V=0.3
Capsular ligaments	E=1	V=0.3

Table 5-20: Nonlinear material properties of ligaments (Kumaresan et al., 1999; Toosizadeh & Haghpanahi, 2011)

ALL		PLL		ISL		LF		CL	
Def. (mm)	Force (N)	Def. (mm)	Force (N)	Def. (mm)	Force (N)	Def. (mm)	Force (N)	Def. (mm)	Force (N)
1.4	35.5	1.0	29.0	1.3	16.9	19	45.9	1.8	53.6
2.7	64.9	2.0	51.4	2.7	24.4	3.7	82.4	3.9	87.9
4.1	89.7	3.0	71.3	4.0	29.5	5.6	119.6	5.8	109.4
5.4	108.6	4.0	85.8	5.4	32.9	7.5	133.7	7.7	125.8
6.8	119.6	5.0	94.7	6.7	34.9	9.4	147.2	9.7	134.8

Table 5-21: Material properties of the spine (Kumaresan et al., 1999)

Cervical structure	Element type	Young Modulus (MPa)	Poisson Ratio
Cortex	Solid	12000	0.3
Cancellous core	Solid	100	0.2
endplate	Solid	600	0.3
Posterior elements	Solid	3500	0.25
Annulus ground substance	Solid	4.7	0.45
Annulus fibers	Rebar	500 (20%)	0.3
Nucleus pulposus	Fluid	1666.7	-
Facet articular cartilages	Solid	10.4	0.4
Facet synovial fluid	Fluid	1666.7	-
Facet synovial membrane	Membrane	12	0.4
Uncovertebral synovial fluid	Fluid	1666.7	-
Uncovertebral synovial	Membrane	12	0.4
Ligaments	Cable	-	-

5.2.2 The range of motion of the upper cervical spine

This Table shows the range of motions of upper cervical in degree (Table 5-22).

Table 5-22: Range of motion some cervical spine from different sources (Zhang et al., 2006)

ALL	Panjabi (1988)		Goel (1988)			Panjabi (2001)		
Level	C0- C1	C1-C2	C0- C1	C1-C2	C5- C6	C0- C1	C1-C2	C5- C6
Moment (Nm)	1.5	-	0.3	-	1.0	1.0	-	1.0
Flexion (deg)	3.5 (0.6)	11.5 (2.0)	6.5 (2.5)	4.9 (2.0)	3.1 (0.9)	7.2 (2.5)	12.3 (2.0)	5.5 (2.6)
Extension (deg)	21 (1.9)	10.9 (1.1)	16.5 (7.6)	5.2 (2.9)	2.8 (0.8)	20.3 (4.6)	12.1 (6.5)	4.4 (2.8)
Axial rotation (deg)	7.9 (0.6)	38.3 (1.7)	2.4 (1.2)	23.3 (11.2)	1.5 (0.5)	4.9 (3.0)	28.4 (4.8)	2.5 (1.0)
Lateral bending (deg)	5.6 (0.7)	4 (0.8)	3.4 (2.8)	4.3 (2.8)	1.6 (0.8)	4.5 (1.5)	3.3 (2.3)	3.3 (1.5)

5.3 Preliminary conclusions

In order to make the model based on the human spine, after getting the anatomical behavior and the model based on anatomical behavior human lumbar spine in the previous chapter, it is necessary to know the mechanical behavior of each part to create the optimal model. Therefore, each part of the human lumbar spine was analyzed and defined in this chapter. Each part has different behavior and different mechanical behavior based on its position and its directions.

There are tables below (Table 5-23, 5-24) that are contained all properties for each part of human lumbar spine which are useful for further study to make the model by the numerical way in virtual space. For modeling based on the mechanical part of the human lumbar spine, we have to change a little mechanical behavior of some parts according to validate data to get the better model by software. Usually, spine materials are non-homogeneity, anisotropy, and nonlinearities, but for making analyze easily they are assumed linear (Zafarparandeh, Erbulut, Lazoglu, & Ozer, 2014), isotropic and with elastic orthotropic structure (Ben-Hatira, Saidane, & Mrabet, 2012; Sahani, 2015), and homogeneous (Tyndyk, Barron, McHugh, & O Mahoney, 2007; Zhang et al., 2006).

Table 5-23: Material parameters for each Part of the Lumbar Spine

component	elements				Material properties					
	Type			Number	formulation	E(MPa)	v	G (MPa)	a	n
Bony endplate	8-noded brick	solid		384	Linear elastic	1000	0.3		0.9	0.7
Cortical shell	8-noded brick	solid	10 node tetrahedral (solid)	672	Linear elastic	12000	0.3	2400	0.9	0.7
Trabecular core	8-noded brick/20-noded brick	solid	10 node tetrahedral (solid)	3936	Linear elastic	344/140/100	0.2	77	0.9	0.7
Bony ring (posterior)	4-noded brick	solid	10 node tetrahedral (solid)	17242-33226	Linear elastic	3500/3000	0.3		0.9	0.7
Cartilage endplate	8-noded brick		wedge	384	Linear elastic	24/10.4	0.4		0.9	0.7
Endplate						1200/500	0.29		0.9	0.7
nucleus	8-noded brick	Fluid	10 node tetrahedral (Hyper elastic solid)	864	Linear elastic	1/2	0.49 9		0.9	0.7
Annual matrix	8-noded brick/ Tension-only link/3d cable	rebar	10 node tetrahedral (solid)	672	Hyperelastic, Mooney-Rivlin	4.2/8/Neo-Hooke	0.45		0.9	0.7
Annulus Fibers	2-noded brick/3d cable	rebar	hexahedron	448		500/550 Area(20% volume fraction)/ Neo-Hooke	0.3		0.9	0.7
Fiber inner			Truss	592		360	0.3			
Fiber layer 2			Truss	592		408	0.3			

Fiber layer 3			Truss	592		455	0.3			
Fiber layer 4			Truss	592		503	0.3			
Fiber outer			Truss	592		550	0.3			
Facet Articular cartilages		solid				10.4	0.4			
Facet synovial fluid		fluid				12				
Facet synovial membrane		membrane				12	0.4			
Uncovertebral synovial fluid		fluid				1666.7				
Uncovertebral synovial membrane		membrane				12	0.4			
Facet joint	Non-linear contact		Tension-only link			3500			0.9	0.7

(Ben-Hatira et al., 2012; Kumaresan et al., 1999; Kuo et al., 2010; Li et al., 2017; Maurel, Lavaste, & Skalli, 1997; Noailly, Lacroix, & Planell, 2003; Sahani, 2015; Salvatore et al., 2018; Schmidt et al., 2006; Toosizadeh & Haghpanahi, 2011; Tyndyk et al., 2007; Wu & Chen, 1996; Zhang et al., 2006)

Table 5-24: Material parameters of Ligaments

	Element					Area		Material properties		
	Type				Number	Transition strain (%)	mm ²	E (MPa)	ν	Young's modulus (MPa)
ALL	3d cable/tension only	wedge	SPRING	10 node tetrahedral (Hyper elastic solid)	1032/9/9046	12	6.1/63.7/53	Non-linear force-deflection response		54.5/10/7.8-20
PLL	3d cable/tension only	wedge	SPRING	10 node tetrahedral (Hyper elastic solid)	258/5/3844	11	5.4/20/16			20/10/10-50
LF	3d cable/tension only	tetrahedron	SPRING	10 node tetrahedral (Hyper elastic solid)	9/6/3042	6.2	50.1/40/67			1.5/10/15-19.5
TL	3d cable/tension only	tetrahedron	T3D2	10 node tetrahedral (Hyper elastic solid)	1/6/6678	18	5/1.8/1.8	54.4	0.3	20/10-59
CL	3d cable/tension only	membrane	T3D2	10 node tetrahedral (Hyper elastic solid)	26/10/3220	25	48.4/46.6/30	24.4	0.3	20/7.5-33
ISL	3d cable/	tetrahedron	T3D2	10 node tetrahedral (Hyper	12/9/2856	20	42.7/13.1/40/26	16.9	0.3	1.5/10/10-11.6

	tension only			elastic solid)						
SS L	3d cable/ tension only	tetrahedron	T3D 2	10 node tetrahedral (Hyper elastic solid)	150/32657	14	38.9/30/23	34.1	0.3	1.5/8-15

(Ben-Hatira et al., 2012; Ha, 2006; Ibarz et al., 2012; Kumaresan et al., 1999; K. K. Lee et al., 2004; Li et al., 2017; Maurel et al., 1997; Toosizadeh & Haghpanahi, 2011; Tyndyk et al., 2007; Zhang et al., 2006)

By making and modeling each geometry in the virtual space numerically, there is the capacity of checking and evaluating that structure easily. So the human lumbar spine as the example part of the human spine is modeled by this method in the next chapter by the anatomical and mechanical properties which are gained from the verified data during these two chapters of four and five.

CHAPTER VI: THE FINITE ELEMENT APPROACH

6.1 Finite element analysis (FEA)

Finite element analysis or the finite element method (FEM) (Picture 6-1) is the product of the digital age after the promotion of digital computers in the 1950s. It is a numerical (Harish, n.d.) and a computer-based analysis tool by using mathematics (Peterson, n.d.) to simulate and analyze engineering products and systems (“Introduction to finite element analysis,” n.d.) in a virtual environment (“Finite Element Analysis (FEA),” n.d.), which is used by engineers and scientists in a wide range of industries such as automotive, biomechanical industries, and aeronautical (“Finite Element Analysis (FEA),” n.d.).

Pic 6-1: Finite element method

The finite element method involves modeling the structure by using small interconnected elements (small regions not separate entities like bricks) called “Elements” which are connected to one another in “Nodes” (like the glue that attaches elements together) (“Finite Element Analysis,” n.d.; “Finite Element Analysis (FEA),” n.d.). The complete set of elements is “Meshes” and this process is “Discretization” or “Meshing” (Peterson, n.d.) (Picture 6-2) (Figure 6-1). The simple equations that model these FE are then assembled into a larger system of equations that models the entire problem (Reddy, 2010).

Fig 6-1: Element and Node definition

Pic 6-2: Element and node definition

Each modeling requires a different range of properties to make such as section areas, thickness, inertia, stiffness, torsional constant, transverse shear, various load (stress-strain,

fluid pressure), nodal (forces, moments, displacements, velocities, accelerations, temperature, heat, sound, and electromagnetic), elemental (distributed loading, pressure, temperature and heat flux), as well as acceleration body loads (gravity) (“FEA software,” n.d.; “Finite Element Analysis (FEA),” n.d.; “Introduction to finite element analysis,” n.d.).

The most important step in FEA is a validation of the model. Testing should include decompression and stabilization procedures simulated in the FEM (Ha, 2006).

6.1.1 Development of FE

The exact date of the invention of the finite element method (FEM), is attributed to the work by Euler, in early 16th century. Then, the earliest mathematical papers on FE analysis are in the works of Schellbach (1851) and Courant (1943) (“Finite Element Analysis,” n.d.; Harish, n.d.) to solve complex elasticity and structural analysis problems in civil and aeronautical engineering. Then it developed by A. Hrennikoff and R. Courant (“The Finite Element Method (FEM),” n.d.) (the early 1940s), Ioannis Argyris, K. Feng (the later 1950s and early 1960s) (Bathe, 2006). In these years, NASA sponsored the original version of Nastran, UC Berkeley made the FE program Sap IV, and Det Norske Veritas (In Norway) developed Sesam in 1969 for use in analysis of ships. Most commercial FEM software packages originated in the 1970s (Abaqus, Adina, Ansys, etc.). An exact mathematical basis of the FE books was provided in 1971 by Zienkiewicz, 1972 Babuska and Aziz (Harish, n.d.), and 1973 with the publication by Strang and Fix. This method book has been generalized for the numerical modeling of physical systems in engineering disciplines, electromagnetism, heat transfer, and fluid dynamics (“Finite Element Analysis,” n.d.)(Bathe, 2006). So, in the early 1960s, engineers used the method for approximate solutions to problems in stress analysis, fluid flow, heat transfer, and other areas. In the late 1960s and early 1970s, the FEM was applied to a wide variety of engineering problems (Bathe, 2006).

6.1.2 Restrictions of FE

The small changes in the parameters lead to a large fluctuation in the solution (Harish, n.d.) because it is a tool and depends on its user.

6.1.3 Advantages of FE

Using the Finite Elements method have these benefits as follow:

- Handling any mathematical or physical problem (Smith, Griffiths, & Margetts, 2013) (“Introduction to finite element analysis,” n.d.) such as linear static structural, non-linear static structural, linear dynamic, non-linear dynamic, linear steady-state thermal, and non-linear thermal (“Introduction to finite element analysis,” n.d.).
- Optimizing product performance and cost.
- Reducing development time.
- Eliminating or reducing testing.

- Being the first-time achievement of the required quality.
- Improving safety.
- Making satisfaction of design codes.
- Improving information for engineering decision making.
- Fuller understanding of components allowing more design that is rational.
- Making Satisfaction of legal and contractual requirements (“Introduction to finite element analysis,” n.d.).
- Having great freedom to select the elements and its functions (“The Finite Element Method (FEM),” n.d.).
- Making complicated geometries with a wide range of engineering problems (Bathe, 2006).
- Being easy to implement.
- Having quality is often higher than the corresponding FDM approach.
- Modeling of structures in bending or twisting.
- Indicating the distribution of stresses and displacements.
- Reducing the number of physical prototypes and experiments to optimize components (“Finite Element Analysis,” n.d.).
- Modifying basis functions to make an accurate calculation without changing mesh (Peterson, n.d.) (Picture 6-3).
- Predicting product performance and reliability.
- Reducing physical prototyping and testing.
- Evaluating different designs and materials.
- Optimizing designs and reducing material usage (“Finite Element Analysis (FEA),” n.d.).

Pic 6-3: Finite element model example

6.1.4 The major stages of FE

The major stages in the creation of any FE model which Baguley and Hose said in 1997 (“Introduction to finite element analysis,” n.d.) are:

- Selecting analysis type.
- Idealizing of material properties.
- Creating model geometry.
- Applying of supports or constraints.
- Applying of loads.
- Optimizing the solution.
- Developing a feel for the behavior of the structure.
- Assessing the sensitivity of the results to approximations of the various types of data,
- Developing an overall strategy for the creation of the model.
- Comparing the expected behavior of the idealized structure with the expected behavior of the real structure (“Introduction to finite element analysis,” n.d.).

6.1.5 The steps to carry out FE

Finite Element Method (FEM) has three steps to solve any Physical problem (Peterson, n.d.) such as:

- **Preprocessing** (Defining problem by FE model): building an FE model by the user. It includes the following steps (Peterson, n.d.):

- Model Designing.
- Discretization or Meshing.
- Selection of elements.
- Geometrical properties.
- Material selection.
- Constraints or Boundary conditions.
- Loads conditions.
- Type of analysis.

- **Processing** (Solving or solution): conducting numerical analysis by computer (Peterson, n.d.). There are two approaches as applied to structural mechanics problems to get a solution.

- Displacement or stiffness method.
- Force or flexibility method.

- **Post processing** (Interpreting the results or reviewing the results): see the result by the user and various operations can be done in the post-processing (Peterson, n.d.).

- Visualization of Deformed structure.
- Displacements magnitude in three directions.
- Nodal stress magnitude.
- Element stress magnitude.
- Check equilibrium.
- The factor of safety.
- Strain energy.
- Natural frequency.
- Amplitude.
- Time history.
- Thermal stress and strain.
- Plot deformed the structural shape.
- Animate dynamic model behavior.
- Produce color-coded temperature plots.

6.1.6 The software for modeling FE

List of popular FEA software's according to its purposes (Peterson, n.d.):

- Preprocessing tools: ANSA, HyperMesh, Patran.
- Processing tools (Solvers): Abaqus, ADINA, Algor, ANSYS, COSMOS, LS-Dyna, MSC, NASTRAN, NISA, Optistruct, RADIOSS, SAP, STARDYNE.
- Post-processing tools: Hyperview, μ ETA.

6.2 The spine finite element

The human spine is a complex structure for protecting the spinal cord and transferring loads from the trunk and head to the pelvis. The motion segment as a basic unit of the spine consists of two vertebrae, the disc that separates them (Tyndyk et al., 2007), ligaments and facet joints to load the bars (Tyndyk et al., 2007).

The investigation of the biomechanical behavior of the human spine can be received in two ways: experimentation or modeling. The experimental studies are based on animal, artificial, and cadaveric spines models while the modeling is mathematical or computer-based.

Finite Element Analysis (FEA) of the spine (a non-invasive and numerical (Zafarparandeh et al., 2014) tool) helps to understand the following (Tyndyk et al., 2007):

- Determining the role of material properties and the behavior of the spine (Aroeira, Pertence, Kemmoku, & Greco, 2017; Toosizadeh & Haghpanahi, 2011; Tyndyk et al., 2007).

- Performing parametric analysis of loadings (stresses in the discs, vertebrae, and ligaments and very detailed motion data) and comparing it to experimental studies (Toosizadeh & Haghpanahi, 2011; Tyndyk et al., 2007; Zhang & Teo, 2008).
- Studying the load transfer mechanisms and load distributions among different spinal components (Fagan, Julian, Siddall, & Mohsen, 2002; Toosizadeh & Haghpanahi, 2011; Tyndyk et al., 2007).
- Assessing the spinal performance of disease, degenerative changes, trauma, aging, surgery, damaged or healthy spine (Tyndyk et al., 2007; Zhang & Teo, 2008).
- Assisting in the design and development of new spinal implants.
- FE models of the spine have advantages over short “specimens” and can contribute to quantitative spinal biomechanical investigations (Tyndyk et al., 2007).
- Handling the various geometries and materials of the prosthesis (Ha, 2006; Zafarparandeh et al., 2014) and examined them if required (Fagan et al., 2002).
- Analyzing the limitation of different treatment options and surgical interventions and allows testing and optimization of any prostheses before it is even manufactured (Zhang & Teo, 2008).
- The specimen cannot be overloaded or damaged but can be modified to reflect any physiological condition (Fagan et al., 2002).

On the other hand, the vivo methods (Experimentation) is the most direct and obvious way to obtain information of spinal biomechanics and performance, but measuring the internal loads of spine are invasive, costly, and time-consuming (Ha, 2006; Toosizadeh & Haghpanahi, 2011). It has a limitation of getting the specimen, having poor bone quality, being infected by a virus (Du et al., 2016), having specimen variability and issues of repeatability (Fagan et al., 2002).

6.2.1 Different types of FE analysis of the spine

The FE analyses are broadly classified into two categories: dynamic and static analysis with different models.

- The models of dynamic study: consisting of a series rigid bodies of vertebrae (as rigid bodies) which are connected by ligaments and discs like springs to predict the large intervertebral response under loads (Zafarparandeh et al., 2014; Zhang et al., 2006).
- The models of static study: having more detailed in the representation of the spinal materials and geometries of either one or two spinal motion segments to predict the internal stresses, strains and biomechanical responses under complex loading conditions. Therefore, it may not able to provide a realistic response to the physical multi-levels of the spinal column (Zafarparandeh et al., 2014; Zhang et al., 2006).

6.2.2 History of Lumbar FE

In 1975, Nachemson made a review of the mechanics of lumbar discs, towards a better understanding of low back pain. Koreska et al. described the fundamentals of the biomechanics of the lumbar spine and its clinical significance.

In 1974, Belytschko et al. developed a numerical model of an intervertebral disc for three-dimensional stress analysis (the vertebrae were idealized as rigid bodies and discs, ligaments, and connective tissues were represented as deformable elements).

In 1979, Berkson et al. studied the mechanical behavior of 42 fresh human cadaver lumbar segments in compression, anterior, posterior, and lateral shear. Hakim and King' used three-dimensional finite element models (FEMs) to analyze the dynamic response of vertebra, with experimental verification (Fagan et al., 2002). Rohlmann et al. made a much-simplified model of the lumbar spine to examine the effects of stresses of the vertebral, end plates, and intervertebral discs (Fagan et al., 2002).

Recently, many investigations have been presented using poroelastic models to study the mechanical behavior of tissues (Wu & Chen, 1996) (Figure 6-2a, 6-2b).

Fig 6-2: (a) Spine elements model, (b) Human spine finite element model

6.3 The approach of making the FE of the human spine

Many finite element (FE) simulations in vivo or in vitro studies are modeled for biomechanical analyses of the lumbar spine up to now. Most of them are modeled based on simplified models such as a quarter of the vertebrae and discs, half of the vertebrae, or a regular shape of lumbar, so this asymmetry in the geometry made undoubtedly the asymmetric responses (Kuo et al., 2010).

As mentioned before, making FE needs three steps of pre-processing, processing, and post-processing.

6.3.1 Pre-processing step

In this step, the FE model is made through three main levels: the geometric representation, the material representation and the boundary conditions (loading, analyzing and validating) (Zafarparandeh et al., 2014).

6.3.1.1 The geometric representation

Lumber of spine contains four major parts: the vertebral bone, the intervertebral disc, the ligaments, and the facet joints.

- Vertebrae (VB) are constructed by the column of the cancellous bone with cortical shell surrounding it.
- Intervertebral disc (IVD) contains the pulposus soft hydrated nucleus in the central, the annulus part in the surrounding of the nucleus, and the end plates (EPs) between the VB and the IVD.
- A pair of Facet Joints (FJs) is located at the posterior side of each VB to guide and steady the movements of each segment.
- There are some ligaments surrounding the VB and the IVD (Wu & Chen, 1996).

Modeling the spine needs a mixed technique (Ibarz et al., 2012) to get the shape of vertebrae, create manually the others parts of the spine and separate them to the different parts for loading. So at first, it needs to get the shape of the vertebrae, then separate vertebrae's parts, make the other tissues, and separate the others parts of spine manually.

In this part, there is the explanation of the ways of making each part of geometry by different software. Sometimes, some of these properties may be useful for one software for modeling.

6.3.1.1.1 Bone (Vertebrae phase 1): Getting the vertebrae shape

Generating the geometric models require specific methodology. The vertebrae geometry can be captured and portrayed by using a variety of methods such as laser scans, computer drawings (CAD), X-rays in multi-vision, biplanar radiographic images, magnetic resonance imaging (MRI), computed tomography (CT) (Aroeira et al., 2017), electro goniometer, video fluoroscopy, NMR, the inclinometer, animal spines (Ibarz et al., 2012), series of images using Slice-O-Matic 4.3 (Ben-Hatira et al., 2012), vertebrae pictures, or getting the proper shapes from websites or databases.

These parts contain three major way of getting vertebra shape:

- CT scan data: the geometry of the lumber of spine is constructed from CT scanning images (sequential sectional cross with the uniform space of 1 mm (Wu & Chen, 1996), 0.6 mm or 0.8 mm (Du et al., 2016; Ha, 2006; Kumaresan et al., 1999; Kuo et al., 2010; Li et al., 2017; Salvatore et al., 2018; Toosizadeh & Haghpanahi, 2011; Tyndyk et al., 2007). Then it imported into MIMICS software ((Materialise Mimics ver. 10.01) with DICOM file format) to develop bone boundaries and make a smoothing procedure (by replacing each node at a slightly different location, without any reduction in node (or element) numbers). Finally, new volumes (shape) of the vertebra are exported to FE

software to do further steps (Li et al., 2017; Salvatore et al., 2018; Toosizadeh & Haghpanahi, 2011).

- Constructing the mesh of vertebrae by reducing its functional unit (two vertebrae without posterior arch) (Maurel et al., 1997): It constructs based on the pictures of different views of vertebrae in modeling software (Figure 6-3) (Picture 6-4) (Fagan et al., 2002; Noailly et al., 2003).

Fig 6-3: Vertebrae Finite element with Mesh model Pic 6-4: Finite element of vertebrae with mesh

- Obtaining the geometry from sources such as BodyParts3D (open-source anatomical database of an adult human male offered by The University of Tokyo (Vikram Subramani, 2016), GRABCAD (online available for education purpose freely (Sahani, 2015)), and trademark Somso QS-15 (a Roland PICZA laser scanner and processed using the programs Dr. Picza 3 and 3D Editor) (Ibarz et al., 2012).

6.3.1.1.2 Bone (Vertebrae phase 2): Vertebrae division

After getting shape or geometry of vertebrae, it is the time of adding the others parts of the spine (bone tissues) and dividing each part of geometry to minor parts, which have the different character and mechanical behavior.

Pic 6-5: Parts of Vertebrae of Lumber

At first, creating the bone tissues by exporting the previous vertebrae geometry to different software such as CAE software (Toosizadeh & Haghpanahi, 2011), SolidWorks (Zulkifli, Ariffin, & Rahman, 2011) Ansys, Abaqus, Patran, HyperMesh (Li et al., 2017; Tyndyk et al., 2007), Autodesk Maya 3D, MeshMixer (Aroeira et al., 2017), 3ds Max, Blender, Matlab,

Mathematica (Toosizadeh & Haghpanahi, 2011), 3-Matic (Salvatore et al., 2018), Solids (Kumaresan et al., 1998), Meshlabs (Vikram Subramani, 2016), Amira (Schmidt et al., 2006), Solidworks (Sahani, 2015), RapidForm, Mimics, and Lightware to model the seamless attachment of the soft bone tissues such as intervertebral discs, ligaments and facet joints (Toosizadeh & Haghpanahi, 2011) manually (Ha, 2006) and get the universal graphics data format (IGES file format) for importing to the software such as Ansys or Abaqus to analyze them (Picture 6-5) (Ha, 2006; Vikram Subramani, 2016).

For simplicity, spinal components were modeled as isotropic and linear except for spinal ligaments (bilinear) (Kuo et al., 2010). So for the curved boundaries of bone, the vertebra components were modeled as isotropic, linear materials, inhomogeneous, and a tetrahedral 20 node quadratic solid element by removing the unnecessary surface, smoothing the irregular surface during the trimming process, and sometimes having cortical bone properties for the whole vertebra (Picture 6-6) (Zulkifli et al., 2011).

Pic 6-6: Different type of vertebrae Finite elements

The vertebrae are composed of six parts. These are the vertebral body, spinous process, transverse process, lamina, pedicle, and facet joints (Zulkifli et al., 2011). In some examples, the volume of each of the vertebrae was split into two sub-volumes of the vertebral body and posterior bony ring, and these subvolumes meshed separately (Tyndyk et al., 2007).

Each part of the vertebra has two layers, of cortical and cancellous bone with the thickness between 0.4 and 0.5 mm for the different region of cervical and range from 1.5 to 2 mm for thoracic (Aroeira et al., 2017; Toosizadeh & Haghpanahi, 2011; Zulkifli et al., 2011). Due to the complexity and low thickness of the posterior part, the cortical bone shell was not modeled separately from the inner portion cancellous bone for this part (Aroeira et al., 2017). The vertebrae were considered rigid bodies; thus, only the exterior surfaces of the vertebrae have meshed elements (Zafarparandeh et al., 2014).

Usually, 8-noded shell elements representing the cortical bone and 20-noded solid elements representing the cancellous bone (Ha, 2006; Ibarz et al., 2012; Zafarparandeh et al., 2014). Sometimes because of the extremely complex geometry of the posterior part of the lumbar vertebrae, Instead of the 8-noded brick elements, 4-noded tetrahedral elements were used (Tyndyk et al., 2007).

There are two element types for spinal models (Picture 6-7):

- Tetrahedral elements: easy to generate on the curvy surfaces but limited to simulate appropriate material properties
- Hexahedral elements (or eight-node “brick” elements): the preferred element type for 3D nonlinear analysis (Zafarparandeh et al., 2014).

Pic 6-7: Cortical and Trabecular parts of Vertebrae in finite element

6.3.1.1.3 Intervertebral disc

The intervertebral discs between two vertebral bodies are created (Tyndyk et al., 2007) as solid volumes to fill the space between two vertebrae (Zafarparandeh et al., 2014). Then it is divided into nucleus pulposus part (0.5–0.8 of the disc volume or 45% of the total disc area (Tyndyk et al., 2007) or 40% of the center of the disc (Aroeira et al., 2017)), annulus ground, superimposed fibers ((like an inverted “V”)) Only a few fibers in the posterior region were represented) (Toosizadeh & Haghpanahi, 2011), the cartilage, and bony endplates (Tyndyk et al., 2007) (Picture 6-8).

Hyperelastic elements were used to model the fluid-like characteristics of the intervertebral disc (Fagan et al., 2002; Toosizadeh & Haghpanahi, 2011).

The main parts of intervertebral disc have characters as follow:

Pic 6-8: The form of vertebrae, end plates, annulus, nucleus places, and annulus fibers form

- The annulus matrix of the intervertebral disc is mesh, 3D, isotropic, 8-noded solid brick elements (Tyndyk et al., 2007), a tetrahedral mesh (Vikram Subramani, 2016), incompressible volume elements (Noailly et al., 2003) to carry only tensile forces.
- The fiber content of annulus is approximately 20% of the matrix volume (Ha, 2006). It is defined by three radial layers and modeled as a composite material consisting of fibers embedded in a ground substance. For the fibers, 3D truss elements, a bi-oriented fiber, tension-only hyperelastic truss elements (Noailly et al., 2003), crossed cable elements, initially slackened which only work in tension, without cable element in the middle of the disc since the nucleus area has no fiber, anterior disk height of 5.5 mm, and a posterior disk height of 3.5 mm (Maurel et al., 1997). They are aligned in layers to form a crisscross pattern placed at an average angle of 40° to the horizontal plane of the disc or from 25° at the periphery of the intervertebral disc, to 45° for the most inner layer (Noailly et al., 2003).
- The nucleus pulposus is simulated as an incompressible material (Noailly et al., 2003), represented by 8-noded hybrid solid elements or 20-noded solid elements (Ha, 2006), a tetrahedral mesh, soft, behaves like a fluid (Vikram Subramani, 2016), with a low Young's modulus and a Poisson's ratio close to 0.5 (Tyndyk et al., 2007).
- The endplates are generated by extruding the NURBS surface patches on the bottom and top surfaces of vertebrae with 20-noded solid elements (Ha, 2006), and a tetrahedral mesh (Picture 6-9) (Vikram Subramani, 2016).

Pic 6-9: Finite element model of intervertebral disc examples

6.3.1.1.4 Ligaments

Seven spinal ligaments with the name of the anterior longitudinal ligament (ALL), posterior longitudinal ligament (PLL), ligamentum flavum (LF), inter-transverse ligament (TL), capsular ligament (CL), inter-spinous ligament (ISL) and supra-spinous ligament (SSL) were considered in the model (Tyndyk et al., 2007) based on their origin and insertion (Zafarparandeh et al., 2014) (Figure 6-4, Picture 6-10).

Fig 6-4: Ligaments places and Numbers in Finite element

Pic 6-10: Finite element of ligaments

They are modeled by using cable elements, carrying force only in tension, with attachments of one node of bone to the other one node, with the cross-sectional area of 1 square millimeter, no compression (Vikram Subramani, 2016), initially slackened, low initial stiffness of the

ligaments, nonlinear behavior, non-linear springs, truss elements (Tyndyk et al., 2007), hypoelastic (Noailly et al., 2003), spring-cable, membrane types, used 3D truss elements, only under tension, only permitted axial force transmission (Zafarparandeh et al., 2014), and all ligaments were defined by using 41 cable elements.

Finally, all the ligaments are modeled tetrahedral except capsular ligaments which are modeled as membrane elements (Ibarz et al., 2012) (Figure 6-5, Picture 6-11). The ligamentum flavum, anterior and posterior longitudinal ligaments have a pre-tension in the neutral position of the spine were set to 14 N, 2 N and 1.5 N for them respectively and the remaining four ligaments 3D truss elements were chosen to simulate (Tyndyk et al., 2007).

Fig 6-5: Finite element model of Lumber 3-4

Pic 6-11: Finite element model of Cervical

6.3.1.1.5 Facet Joints

Facet joints have complex structures, which contain joint capsules, cartilage, the synovial membrane, and a cavity filled with a synovial fluid lubricant, which reduces friction between articular cartilages to levels close to zero. This joint structure is uniaxial and only allows gliding movement (Aroeira et al., 2017; Zafarparandeh et al., 2014) (Picture 6-12). The space between the two cartilages is usually defined by sliding or gap elements and used a squeeze-film-bearing model to represent the synovial fluid and hexagonal elements for simulating the articular cartilage (Zafarparandeh et al., 2014).

Pic 6-12: Finite element model of facet joints

The facet joints as a small anatomic structure, approximately 1 cm high, exhibit smooth concavity (Aroeira et al., 2017) were modeled as a 3D frictionless (Tyndyk et al., 2007), cartilages, hypo elastic, nearly incompressible and frictionless contact body, non-linearity (Noailly et al., 2003), the elastic properties, coefficient of friction of 0.1, modelled by contact pair surface elements (the effect of the capsular ligaments (Kuo et al., 2010)), a refinement of 0.3 mm (Aroeira et al., 2017), gap elements to contact between the articulating face, this initial gap (nodes with zero or negative distances) about 0.8 mm and an average of 25 gap elements were used to model each facet joint contact (Tyndyk et al., 2007) (Picture 6-13).

Pic 6-13: Finite element model of human spine examples

6.3.1.2 Material representation

This section contains the material properties and boundary conditions (Kumaresan et al., 1998) with software such as Abaqus, Ansys. After modeling, numerical methods with its properties are used to determine force-displacement (stress-strain) behavior (Toosizadeh & Haghpanahi, 2011). These properties are based on chapter five (mechanical behavior) and what we described there (Table 6-1, 6-2).

Usually, spine materials are non-homogeneity, anisotropy, and nonlinearities, but for making analyze easily they are assumed linear (Zafarparandeh et al., 2014), isotropic, with elastic orthotropic structure (Ben-Hatira et al., 2012; Sahani, 2015), and homogeneous (Tyndyk et al., 2007; Zhang et al., 2006).

Table 6-1: Material parameters for each Part of the Lumbar Spine

Component	Elements				Material properties					
	Type			Number	Formulation	E(MPa)	v	G (MPa)	a	n
Bony endplate	8-noded brick	solid		384	Linear elastic	1000	0.3		0.9	0.7
Cortical shell	8-noded brick	solid	10 node tetrahedral (solid)	672	Linear elastic	12000	0.3	2400	0.9	0.7

Trabecular core	8-noded brick/20-noded brick	solid	10 node tetrahedral (solid)	3936	Linear elastic	344/140/100	0.2	77	0.9	0.7
Bony ring (posterior)	4-noded brick	solid	10 node tetrahedral (solid)	17242-33226	Linear elastic	3500/3000	0.3		0.9	0.7
Cartilage endplate	8-noded brick		wedge	384	Linear elastic	24/10.4	0.4		0.9	0.7
Endplate						1200/500	0.29		0.9	0.7
nucleus	8-noded brick	Fluid	10 node tetrahedral (Hyper elastic solid)	864	Linear elastic	1/2	0.499		0.9	0.7
Annual matrix	8-noded brick/ Tension-only link/3d cable	rebar	10 node tetrahedral (solid)	672	Hyperelastic, Mooney-Rivlin	4.2/8/Neo-Hooke	0.45		0.9	0.7
Annulus Fibers	2-noded brick/3d cable	rebar	hexahedron	448		500/550 Area(20% volume fraction)/ Neo-Hooke	0.3		0.9	0.7
Fiber inner			Truss	592		360	0.3			
Fiber layer 2			Truss	592		408	0.3			
Fiber layer 3			Truss	592		455	0.3			
Fiber layer 4			Truss	592		503	0.3			
Fiber outer			Truss	592		550	0.3			
Facet Articular cartilages		solid				10.4	0.4			
Facet synovial fluid		fluid				12				
Facet synovial membrane		membrane				12	0.4			
Uncovertebral synovial fluid		fluid				1666.7				
Uncovertebral synovial membrane		membrane				12	0.4			
Facet joint	Non-linear contact		Tension-only link			3500			0.9	0.7

(Ben-Hatira et al., 2012; Kumaresan et al., 1999; Kuo et al., 2010; Li et al., 2017; Maurel et al., 1997; Noailly et al., 2003; Sahani, 2015; Salvatore et al., 2018; Schmidt et al., 2006; Toosizadeh & Haghpanahi, 2011; Tyndyk et al., 2007; Wu & Chen, 1996; Zhang et al., 2006)

Table 6-2: Material parameters of Ligaments

	Element					Area		Material properties		
	Type				Number	Transition strain (%)	mm ²	E(MPa)	v	Young's modulus(MPa)
ALL	3d cable/ tension only	wedge	SPRINGA	10 node tetrahedral (Hyper elastic solid)	1032/9/9046	12	6.1/63.7/53	Non-linear force-deflection response		54.5/10/7.8-20
PLL	3d cable/ tension only	wedge	SPRINGA	10 node tetrahedral (Hyper elastic solid)	258/5/3844	11	5.4/20/16			20/10/10-50

LF	3d cable/ tension only	tetrahe dron	SPRINGA	10 node tetrahedral (Hyper elastic solid)	9/6/304 2	6.2	50.1/40/6 7			1.5/10/15-19.5
TL	3d cable/ tension only	tetrahe dron	T3D2	10 node tetrahedral (Hyper elastic solid)	1/6/667 8	18	5/1.8/1.8	54.4	0.3	20/10-59
CL	3d cable/ tension only	membr ane	T3D2	10 node tetrahedral (Hyper elastic solid)	26/10/3 220	25	48.4/46.6/ 30	24.4	0.3	20/7.5-33
ISL	3d cable/ tension only	tetrahe dron	T3D2	10 node tetrahedral (Hyper elastic solid)	12/9/28 56	20	42.7/13.1/ 40/26	16.9	0.3	1.5/10/10-11.6
SSL	3d cable/ tension only	tetrahe dron	T3D2	10 node tetrahedral (Hyper elastic solid)	150/32 657	14	38.9/30/2 3	34.1	0.3	1.5/8-15

(Ben-Hatira et al., 2012; Ha, 2006; Ibarz et al., 2012; Kumaresan et al., 1999; K. K. Lee et al., 2004; Li et al., 2017; Maurel et al., 1997; Toosizadeh & Haghpanahi, 2011; Tyndyk et al., 2007; Zhang et al., 2006)

There is some suggestion of features for each part of the spine for representing materials of them. The properties, which are useful for different modeling, depend on the aim of the study and the software, which are chosen.

6.3.1.2.1 Vertebrae

To assign the material properties of vertebrae, the vertebral mesh is divided into **four** different groups based on the mineral density of the bone. These groups are cortical bone, cancellous bone, the posterior part and endplates (Ibarz et al., 2012; Zafarparandeh et al., 2014) (Picture 6-14)

Pic 6-14: Finite element of lumbar vertebrae examples

- Cortical bone: rigid bodies (Ibarz et al., 2012; Zafarparandeh et al., 2014), the soft tissue (Ibarz et al., 2012; Zafarparandeh et al., 2014), porous (The fluid contents are low), elastic (Wu & Chen, 1996; Zafarparandeh et al., 2014), orthotropic solids, isotropic (Toosizadeh & Haghpanahi, 2011), linear (Kuo et al., 2010), isoparametric, eight-nodded solid elements (Kumaresan et al., 1999; K. K. Lee et al., 2004; Zafarparandeh et al., 2014),

solid tetrahedral linear elements (to preserve the original geometry of bone) (Kuo et al., 2010), with the compressibility of 0.9 (Wu & Chen, 1996), and thickness of 0.5–2.5 mm with mesh elements of high quality (Du et al., 2016) or 1.0 mm thick (K. K. Lee et al., 2004) or 0.35 mm thick or is modelled as triangular shell elements (M3D3) with element side in the range of 1 mm to 3 mm (Kuo et al., 2010), or with a thickness of 0.4mm (Li et al., 2017).

- Cancellous bone: rigid bodies, the soft tissue (Ibarz et al., 2012; Zafarparandeh et al., 2014), porous (The fluid contents are low), elastic (Wu & Chen, 1996; Zafarparandeh et al., 2014), orthotropic solids, isotropic (Toosizadeh & Haghpanahi, 2011), linear (Kuo et al., 2010), isoparametric, eight-nodded solid elements (Kumaresan et al., 1999; K. K. Lee et al., 2004; Zafarparandeh et al., 2014), solid tetrahedral linear elements (to preserve the original geometry of bone) (Kuo et al., 2010), and the compressibility of 0.9 (Wu & Chen, 1996).

- The posterior part: rigid bodies, the soft tissue (Ibarz et al., 2012; Zafarparandeh et al., 2014), porous (The fluid contents are low), elastic (Wu & Chen, 1996; Zafarparandeh et al., 2014), orthotropic solids, isotropic (Toosizadeh & Haghpanahi, 2011), linear (Kuo et al., 2010), isoparametric, eight-nodded solid elements (Kumaresan et al., 1999; K. K. Lee et al., 2004; Zafarparandeh et al., 2014), and solid tetrahedral linear elements (to preserve the original geometry of bone) (Kuo et al., 2010).

- Endplates: rigid bodies, the soft tissue (Ibarz et al., 2012; Zafarparandeh et al., 2014), porous (The fluid contents are low), elastic (Wu & Chen, 1996; Zafarparandeh et al., 2014), orthotropic solids, isotropic (Toosizadeh & Haghpanahi, 2011), linear (Kuo et al., 2010), isoparametric, eight-nodded solid elements (Kumaresan et al., 1999; K. K. Lee et al., 2004; Zafarparandeh et al., 2014), solid tetrahedral linear elements (to preserve the original geometry of bone) (Kuo et al., 2010), and the thickness (were assumed at the superior and inferior surfaces of the intervertebral disc (Li et al., 2017) as triangular shell elements of 0.5 mm thick or in the range of 1 mm to 3 mm (Kuo et al., 2010), or with a thickness of 0.4mm (Li et al., 2017).

6.3.1.2.2 Intervertebral disc

Each intervertebral disc was modeled as a central nucleus surrounded by an annular ground substance reinforced by fibers (Kumaresan et al., 1999) acting at approximately ± 30 degree from the transverse plane. The upper and lower surfaces of each disc were covered with the end plate (Du et al., 2016).

- Nucleus pulposus: it is composed of a fluid material (Zafarparandeh et al., 2014) like a non-compressible fluid (Ibarz et al., 2012), hyper elastic Mooney-Rivlin (Ibarz et al., 2012) isotropic elastic model (Zafarparandeh et al., 2014), linearly elastic (Salvatore et al., 2018), hyper elastic Mooney–Rivlin material properties (The fluid-like behavior) with two parameters (C1, C2) (Toosizadeh & Haghpanahi, 2011), near incompressible tetrahedral elements to preserve the original geometry (Kuo et al., 2010), 48% volume of the entire disc volume (Du et al., 2016), the porosity (n) of 0.70 (Wu & Chen, 1996) or 0.5 (K. K. Lee et al., 2004), compressibility of 0.9 (Wu & Chen, 1996), incompressible,

3-D brick elements, and Young's modulus of 1MPa to simulate hydrostatic character (K. K. Lee et al., 2004). Some are modeled as isotropic and some are modeled as Mooney-Rivlin hyper elastic materials (Vikram Subramani, 2016).

- Annulus ground substances: a matrix of ground substance is a composite material which (K. K. Lee et al., 2004) is reinforced with collagen fibers, isotropic elastic model (Zafarparandeh et al., 2014), a nonlinear only tension behaviour (Ibarz et al., 2012), a Neo-Hookean solid (Salvatore et al., 2018), hyper elastic, Mooney–Rivlin material properties (The fluid-like behavior) with two parameters (C_1 , C_2) (Toosizadeh & Haghpanahi, 2011), solid tetrahedral linear elements to preserve the original geometry (Kuo et al., 2010) with the porosity (n) of 0.70, compressibility of 0.9 (Wu & Chen, 1996) as a net structure with comprised of 20% of the ground substance volume (Li et al., 2017), consist of three (seven (Salvatore et al., 2018)) consecutive laminar layers which were oriented on average at to the endplate (K. K. Lee et al., 2004) approximately $\pm 30^\circ$ (Vikram Subramani, 2016) or $\pm 24^\circ$ to (Schmidt et al., 2006) the horizontal plane (Salvatore et al., 2018), and some are modeled as Mooney-Rivlin hyper elastic materials (Vikram Subramani, 2016).

- Annulus fibrosus (collagen fibers (Zafarparandeh et al., 2014)): 3D (two-node truss elements (Salvatore et al., 2018) cable elements that sustained tension only (K. K. Lee et al., 2004) varies from 23% in the outermost layer to 5% in the innermost layer (Salvatore et al., 2018). 20% of the annulus volume is allocated to these fibers (Toosizadeh & Haghpanahi, 2011) which are arranged in a crisscross manner with 25 degrees orientation (Kumaresan et al., 1999), hyper elastic, Mooney–Rivlin material properties (The fluid-like behavior (Schmidt et al., 2006)) with two parameters (C_1 , $C_2 = 0.56, 0.14$) (Toosizadeh & Haghpanahi, 2011), solid tetrahedral linear elements to preserve the original geometry (Kuo et al., 2010), the fiber-reinforced concrete (Kumaresan et al., 1999), nonlinear tension-only links, with 1.2×10^6 cross-section area, initial strain about -0.1 pre-stress in the fibers (Toosizadeh & Haghpanahi, 2011), with The porosity (n) of 0.70, compressibility of 0.9 (Wu & Chen, 1996), the fibers were modelled using a rebar definition (Du et al., 2016), rebar elements to carry only tensile forces and no-compression (Salvatore et al., 2018), solid elements (Kumaresan et al., 1999), homogenous, eight fiber layers were defined with a crosswise pattern (Schmidt et al., 2006), Young's modulus of 4.2 MPa., non-linear function, and external lamellae behave stiffer than internal lamellae (outermost layers 1–2: 1.0, layers 3–4: 0.9, layers 5–6: 0.75, innermost layers 7–8: 0.65) (Schmidt et al., 2006).

- Endplate: The upper and lower surfaces of each disc were covered with the end plate of 0.6 mm (Du et al., 2016), or 0.5mm thickness (Zafarparandeh et al., 2014) or less than 0.7 mm (Maurel et al., 1997) with incompressible fluid elements (Kumaresan et al., 1999).

The uncovertebral joints located bilaterally in the two intervertebral discs were also modeled using incompressible fluid elements, and the synovial membranes enclosing the uncovertebral joints were modeled using membrane elements (Picture 6-15) (Kumaresan et al., 1999).

Pic 6-15: Finite element model of intervertebral disc examples

6.3.1.2.3 Ligaments

The role of ligaments is to resist tensile or distractive forces (Zafarparandeh et al., 2014). Spine ligaments as mentioned before are the anterior longitudinal (ALL), posterior longitudinal (PLL), ligament flavum (LF), capsular ligaments (CL), interspinous ligament (ISL), supraspinous ligament (SSL), and transverse ligaments (TL) (Du et al., 2016; Kumaresan et al., 1999; K. K. Lee et al., 2004).

Ligaments are tension-only, truss elements (Li et al., 2017), an axial linear spring element, with a constant of 15 N/mm correctly applied to nodes (Wu & Chen, 1996), nonlinear tension-only springs, with a pre-stress of -0.1 strain, both geometrical and material nonlinearity (Toosizadeh & Haghpanahi, 2011), 3D nonlinear spring elements, nonlinearly in tension only (Du et al., 2016), bilinear materials, narrow strip-shaped membrane elements (M3D3), no resistance in compression (Kuo et al., 2010), linear elastic material, no compression option (Vikram Subramani, 2016), 3D two node truss elements, nonlinear hypo elastic material properties (Salvatore et al., 2018), nonlinear tension-active cable elements (Kumaresan et al., 1999) as simple elastic beam elements, tension-only cable elements are preferred (Zafarparandeh et al., 2014), nonlinear stress-strain behavior opposite all material properties (K. K. Lee et al., 2004), nonlinear, and only tension behavior, as a bilinear model in the strain range (Ibarz et al., 2012).

6.3.1.2.4 Facet Joints

To contact the gap between vertebrae, facet joints are introduced between respective nodes of adjacent facets, sliding parallel to the surface of the facets perpendicularly to the plane of these facets (Maurel et al., 1997).

Facet joints take the same material properties as for the cortical shell (Wu & Chen, 1996), frictionless 3D surface-to-surface, soft contact, an initial gap of 0.5 mm (Du et al., 2016) the compressibility 0.9 (Wu & Chen, 1996).

Facet joint contains the synovial fluid, articular cartilages, synovial membrane (Kumaresan et al., 1999) (Du et al., 2016).

- The synovial fluid: incompressible fluid elements.
- Articular cartilages: solid elements with a thickness of 0.5 mm (Du et al., 2016).
- Synovial membrane: membrane elements (Kumaresan et al., 1999).

Facet joints play an important role in mechanical behavior, so must be included in the analysis (Wu & Chen, 1996).

6.3.1.3 The boundary and loading conditions

For more complicated models, it is more convenient to use FE software such as Abaqus (Ibarz et al., 2012; Toosizadeh & Haghpanahi, 2011; Tyndyk et al., 2007) and Ansys for setting up the flexion-extension and compression-tension simulations (Li et al., 2017; Vikram Subramani, 2016), calculating the geometry of the vertebrae (Ben-Hatira et al., 2012; K. K. Lee et al., 2004; Maurel et al., 1997; Toosizadeh & Haghpanahi, 2011; Zulkifli et al., 2011) with its mesh function, and conducting the probabilistic analysis (Zulkifli et al., 2011).

6.3.1.3.1 Loading

The loads were applied to the upper vertebral body (Wu & Chen, 1996) or node on the superior surface of upper vertebral body (Vikram Subramani, 2016) (axial torque, moments of flexion, extension, lateral flexion, compression, rotation) (Noailly et al., 2003; Zafarparandeh et al., 2014) and the lower base of the lower vertebral body being embedded and fixed as a rigid plate (zero displacements) (Picture 6-16a, 6-16b) (Ben-Hatira et al., 2012; Du et al., 2016; Ha, 2006; Kumaresan et al., 1999; Kuo et al., 2010; Maurel et al., 1997; Noailly et al., 2003; Schmidt et al., 2006; Tyndyk et al., 2007; Zafarparandeh et al., 2014).

Pic 6-16: (a) Definition of Each loads of Human spine, (b) How to Load bars on vertebrae

According to picture 6-16a, there are (a) three planes of movements with different position of movements such as (b) compression (c) lateral Bending (d) torsion (e) flexion (f) extension (Gómez, Lorza, Bobadilla, & García, 2017).

Based on picture 6-16b, the way of loading the bars for each type of body position are (a) without load (b) compression load (c) flexion load (d) lateral bending load (e) torsion load (f) Shear load (Gómez et al., 2017).

There are three types of loading bars such as pure compression, pure moment, and pure compression with a pure moment. The number of bars for each type of loading:

- Under pure moment, only the pure moment of 7.8Nm (Noailly, Wilke, Planell, & Lacroix, 2007) or 7.5 Nm (Tyndyk et al., 2007).
- Under pure compression, only the compression of 1000N (the load of standing or walking situation for a 70 kg person) (Campbell, Coombs, Rao, Rullkoetter, & Petrella, 2016; Rao, 2012; Xu, Yang, Lieberman, & Haddas, 2017), or 1200 N (upper body weight) (Salvatore et al., 2018) or compression of 400 *nn* (the pre compression due to the body weight) (Ibarz et al., 2012; Sahani, 2015).
- Under pure compression and pure moment, the mixed of compressive force and moment are applied (Shirazi-Adl (1994)) (Xu et al., 2017) (Table 6-3).

Table 6-3: The number of Loads

Body position	Compressive force (N)	Moment (Nm)	References
Flexion	1175	7.5	Rohlmann et al.(2009)
Extension	500	7.5	Rohlmann et al.(2009)
Lateral bending	700	7.8	Dreischarf et al.(2012)
Axial Rotation	720	5.5	Dreischarf et al.(2011)

6.3.2 Processing part

This part contains arranging the solutions for the FE model to get its data according to the goals of the study.

The normal solutions for these type of study are stress curves, strain curves, and deformation data. More than these data based on the verified data for this type of study, we have to get the external and internal response of human lumbar spine too. These data are the ROM (range of motion) which is the number of differences between two vertebrae before and after of loading bars for external response and IDP which is the amount of the pressure in the nucleus for an internal response which are the parameters chosen for validation from these experimental resources (Campbell et al., 2016; Xu et al., 2017).

6.3.3 Post-processing part

At the post-processing step, we are going to analyze the model data for validating it to do further research.

6.3.3.1 Validating

There are high variability of tissue material properties, the anatomical spinal structures, and different loading and boundary conditions for vivo and vitro dataset to make FE modeling (Dreischarf et al., 2014), due to the range of differences among the age, gender, and degeneration of the human spine. Recently, Dreischarf et al. (2014) reported that combined several FE models may better represent the predictive capability of the modeling method than simulation results from one deterministic model (Xu et al., 2017). But, needing standards to get the confidence of modeling of FE models based on these three main steps: code verification, and task-specific validations of the model (Dreischarf et al., 2014; Xu et al., 2017) by comparing with experimental data and checking its sensitivity to the larger size.

- Code Verification focuses on the appropriate of the formula and techniques which are used to calculate the model (Xu et al., 2017). It is the least concern because of the vast computational studies (Dreischarf et al., 2014).
- Validation is used to indicate that model is acceptable by comparing with the calculated results from the limited data in vitro or vivo studies (these datasets have a little difference in various studies, but these differences are often neglected) (Dreischarf et al., 2014; Noailly et al., 2007; Xu et al., 2017) or cadaveric studies (Zafarparandeh et al., 2014) for the flexion-extension, axial rotation, and lateral bending moments directions (Du et al., 2016; Ha, 2006; Kumaresan et al., 1999; Zafarparandeh et al., 2014).
- Sensitivity is checked to understand if this structure is modeled well. This sensitivity is an analysis based on the goal of studying.

6.3.3.1.1 Code verification

There are the vast of data that are explained in chapter five and the Table 6-1 and Table 6-2 are gained from these reliable sources.

6.3.3.1.2 Validation method

For validation, FE models were tested in loading conditions of combined loading modes consisting of both pure moment and pure compression (Table 6-3). The simulation results were compared with in vivo experimental data (direct or indirect validation) by Rohlmann et al. 2009 and in vitro experimental results by Percy et al. 1984; Percy & Tibrewal 1984; Percy 1985; Wilke et al. 2001, Dreischarf et al. 2014 (Campbell et al., 2016; Xu et al., 2017).

- Indirect validation involves comparing FE output with a range of experimental data from multiple specimens.
- Direct validation involves developing an FE model of a specific specimen and comparing FE results to experimental data from the same specimen (Campbell et al., 2016).

There are two main groups of experimental data, which are famous for validation.

- Group 1:

1- For ROM, the median and range of IRA were shown in all four loading modes (Diagram 6-1). The ranges of IRA may be different from the ranges of in vivo experimental data due to the differences in material properties and modeling method in the existing literature (Xu et al., 2017).

- The “Literature review” by Dreischarf et al. 2014.
- The “Previous study” by Ming Xu, James Yang, Isador H. Lieberman, and Ram Haddas, 2016.
- The “vivo data” by Percy et al. 1984, Percy & Tibrewal 1984, and Percy 1985 (Dreischarf et al., 2014; Xu et al., 2017).

2- For IDP, the median and range of simulation results were shown in all four loading modes. The median values of IDP in the current study and the existing literature were consistent with the in vivo experimental results at the L4-L5 spinal level under all four loading modes (Xu et al., 2017) (Diagram 6-2).

- The “Literature review” by Dreischarf et al. 2014.
- The “Previous study” by Ming Xu, James Yang, Isador H. Lieberman, and Ram Haddas, 2016.
- The “vivo data” by Wilke et al. 2001 (Dreischarf et al., 2014; Xu et al., 2017).

Diag 6-1: ROM in all spinal level among in vivo and vitro experimental data in flexion (a), Extension (b), lateral bending (c), and axial rotation (d).

Diag 6-2: IDP in all spinal level among in vivo and vitro experimental data under flexion (a), Extension (b), lateral bending (c), and axial rotation (d).

- Group 2:

1- For ROM, there are the data under pure moment and compressive loading (For upper body flexion, a compression force of 1175 N) and intervertebral rotations for six FE models and median in vivo values (with its range) in flexion(a), extension(b), lateral bending(c) and axial rotation(d) (Dreischarf et al., 2014) (Diagram 6-3).

Diag 6-3: intervertebral rotations for six finite element models and median in vivo values (with its range) in flexion(a), extension(b), lateral bending(c) and axial rotation(d)

- Vitro data by Ayturk and Puttlitz, 2001, Kiapour et al., 2012a, Little et al., 2008, Liu et al., 2011, Park et al., 2013, Schmidt et al., 2012, Shirazi-Adl, 1994b, and Zander et al., 2009.

- Vivo data by Percy et al. 1984, Percy & Tibrewal 1984, and Percy 1985.

2- For IDP, Intradiscal pressures for six FE models and median in vivo values (with its range by Wilke et al (2001)) in flexion(a), extension(b), lateral bending(c) and axial rotation(d) (Dreischarf et al., 2014) (Diagram 6-4).

- Vitro data by Ayturk and Puttlitz, 2001, Kiapour et al., 2012a, Little et al., 2008, Liu et al., 2011, Park et al., 2013, Schmidt et al., 2012, Shirazi-Adl, 1994b, and Zander et al., 2009

- Vivo data by Wilke et al. 2001

Diag 6-4: Intradiscal pressures for six finite element models and median in vivo values (with its range by Wilke et al.(2001)) in flexion(a), extension(b), lateral bending(c) and axial rotation(d)

6.3.3.1.3 Sensitivity analysis

This type of analysis is done to check the sensitivity of the model according to the goal of the study. This type of analysis sometimes is about mesh sensitivity, sometimes is material sensitivity. In this study, the scaling sensitivity is checked to test the model for the bigger size.

According to the scale sensitivity analysis, the model is scaled but the equivalent stress, the number of element and node and the material properties don't change. So the amount of pure compression model.

6.4 Preliminary conclusions

According to what explained during these chapters up to now, this chart was made to develop the model based on the FEM. This model belonged to the two motion segments of human lumbar spine L3/L4 under loads of both pure compression and pure moment in four directions of flexion, extension, lateral bending, and axial rotation. This model was checked about its ROM, IDP, stress, and strain curve with experimental data and was scaled to check its sensitivity of that model in bigger size.

The process to develop this model was explained in summary in the chart below (Picture 6-17).

Pic 6-17: The process to develop the human spine by Finite Element method

CHAPTER VII: MODELING THE HUMAN LUMBAR SPINE BY FEM

7.1 Modeling the two motion segments of the spine by FEM

According to what we analyze and describe up to now, spine has three major parts of vertebrae such as cervical, thoracic, and lumbar. Each part has a specific character:

- Cervical: it is much more mobile and flexible than the other spinal regions and it is supporting only the weight of the head (an average of 15 pounds).
- Thoracic: this part has little movements because of thin intervertebral discs and its attachment to the ribs, but it is too strong to protect the vital organs.
- Lumbar: the vertebrae and the spinal canal in this area are the largest of the whole spine. Most of our weight bearing and body movement takes place in this part.

So the lumbar part of the spine was chosen because most of the weight bearing and body movement take place in this part so its characters are beneficial for modeling the buildings structures. The two-motion segment is enough for modeling and expanding it to the whole parts of the lumbar spine due to having the complete character of the whole spine.

7.1.1 Pre-processing step

This part contains three steps of making geometry, representing materials, and loading boundary and conditions.

7.1.1.1 Modeling steps of two motion segments

Modeling the two motion segments according to the previous part needs some steps and some software that is explained in the following. For each step, first, the software is explained, and then the way of the promotion of the model in that software is described.

7.1.1.1.1 Step one (Computed tomography (CT))

The first step of modeling is getting the vertebrae based on CT scan data (the exact and easy way of getting vertebrae geometry).

Computed tomography (CT or CAT scan) is a noninvasive diagnostic imaging procedure to produce horizontal, or axial images (often called slices) in multiple planes, and can generate three-dimensional images ("Computed Tomography (CT or CAT) Scan of the Spine," n.d.) inside of the body by using X-rays and a computer in a separate control room ("Computed Tomography (CT) - Spine," n.d.). It allows the user to see inside the object such as bones, muscles, fat, and organs without cutting (Herman, 2009) on a computer monitor or printed form ("Computed Tomography (CT) - Spine," n.d.) (Picture 7-1). It is an important tool in medical field from the 1970s and it was used for screening the disease over the last two decades in many countries (Burk, 2007).

Pic 7-1: CT scanner

CT scanning works very much like other x-ray examinations by absorbing the x-rays in varying degrees (“Computed Tomography (CT) - Spine,” n.d.), but it can provide more detailed information about the vertebrae, other spinal structures and tissues than standard X-rays (2D). (“Computed Tomography (CT or CAT) Scan of the Spine,” n.d.; Grossman, Neil & Moloney Johns, Amanda, PA-C, MPAS, n.d.) (Picture 7-2a).

The CT scanner is a large, box-like machine with a whole or short tunnel in its center. It scans things, which are lied on a narrow examination table that is sliced into and out of this tunnel, and the machine is rotating around you with its x-ray tube and electronic x-ray detectors. Sometimes, the examination table is moving during the scan, so that the x-ray beam follows a spiral path.

- Advantage

- Modern CT scanners are so fast and easy to scan large sections of the body in just a few seconds.
- CT scanning is painless, noninvasive and accurate.
- CT can be performed if you have an implanted medical device.
- CT scanning of the spine is also performed to:

Diagnosing headache when neuroimaging is indicated (“Computed Tomography (CT) - Spine,” n.d.).

Happening of complex fractures due to injury (Buckwalter, Rydberg, Kopecky, Crow, & Yang, 2001).

In an emergency such as hemorrhage, spinal column damage, stroke, or traumatic brain injury to reveal internal injuries and bleeding quickly enough to help save lives (“Computed Tomography (CT) - Spine,” n.d.).

Evaluating the spine before and after surgery (“Computed Tomography (CT) - Spine,” n.d.; Grossman, Neil & Moloney Johns, Amanda, PA-C, MPAS, n.d.).

Measuring the bone density of bone, most soft tissues, and congenital anomalies.

Detecting various types of tumors in the vertebral column.

Diagnosing herniated intervertebral disk (“Computed Tomography (CT) - Spine,” n.d.).

Displaying three-dimensional imaging of bone, soft tissue and blood vessels at the same time (“Computed Tomography (CT) - Spine,” n.d.).

Making it a good tool for guiding in procedures such as needle biopsies and needle aspirations in many areas of the body due to its real-time imaging (“Computed Tomography (CT) - Spine,” n.d.).

Pic 7-2: (a) CT scan presentation, (b) CT scan examples

- Disadvantage

- CT is a cost-effective imaging tool for a wide range of clinical problems.
- CT is less sensitive to patient movement than MRI (“Computed Tomography (CT) - Spine,” n.d.).
- The radiation used in CT scans damage body cells, including DNA molecules, which can lead to cancer. A person's age plays a significant role in the subsequent risk of cancer. CT scans have 100 to 1,000 time’s higher dose than conventional X-rays (Redberg & Smith-Bindman, 2014), but X-rays used in CT scans have no immediate side effects (“Computed Tomography (CT) - Spine,” n.d.).
- X-ray, MRI, or physical exam is not available or does not provide enough information (Grossman, Neil & Moloney Johns, Amanda, PA-C, MPAS, n.d.).
- CT scanning is not recommended for pregnant women unless medically necessary.
- Mothers should not breastfeed their babies for 24-48 hours after contrast medium CT scan is given.

- The risk of serious allergic reaction to contrast materials that contain iodine is extremely low.
- Children are more sensitive to radiation, so they should have a CT exam only if it is essential with low-dose technique.
- A large person and overweight of 450 pounds can't have CT scan because of the ability to move the table.
- Spinal CT does not consistently show enough detail of the spinal cord and MRI may be more suitable for demonstrating injured ligaments, the status of the intervertebral disks, spinal cord abnormalities and hematomas in the area of the spine ("Computed Tomography (CT) - Spine," n.d.) (Picture 7-2b).

- Getting the CT scan of lumbar

The CT scan of this research was given from the website, <https://www.embodi3d.com/files/category/47-spine-cts/>. This CT scan data is contained of 3 mm slices of normal human lumbar spine without any disease and disability in NRRD format (Picture 7-3).

Pic 7-3: CT scan of L3/L4 for this Thesis

CT scan data are in different types of format such as STL, NRRD, and DICOM file format. According to the file format for going to the next step, the CT scan data was converted into the DICOM format. Some software such as 3D SLICER is good for converting the CT scan data in NRRD format to the DICOM format.

7.1.1.1.2 Step two (3D Slicer 4.8.1)

3D Slicer is a free, open source, flexible, and modular platform (Pieper, Halle, & Kikinis, 2004) software for medical image informatics, processing, visual ling three-dimensional of multi-organ (from head to toe), and researching in image-guided therapy ("Slicer 4.8 released," n.d.; "What is Slicer?," n.d.). It is not approved for clinical use and intended for research ("What is Slicer?," n.d.) (Picture 7-4).

Pic 7-4: Logo of 3D Slicer

3D Slicer was a master's thesis project between the Surgical Planning Laboratory at the Brigham and Women's Hospital and the MIT Artificial Intelligence Laboratory in 1998. Then it was evolved and received up to now. It is still under development (Fedorov et al., 2012) (Picture 7-5).

Pic 7-5: 3D Slicer Example

- Advantage

- Available on multiple operating systems: Linux, MacOSX, and Windows.
- Constructing and visualizing collections of MRI (Hata et al., 2007), CT, US, nuclear medicine, and microscopy data (“What is Slicer ?,” n.d.).
- Supporting MINC image files, 4D NRRD file as segmentation, and DICOM file.
- Volume rendering.
- Displaying arbitrarily oriented image slices (Pieper, Lorensen, Schroeder, & Kikinis, 2006).
- Having powerful plug-in capabilities for adding algorithms and applications (“What is Slicer ?,” n.d.).
- Improving the model by adding new effects such as Grow from seeds, Fill between slices, Surface cut, Mask Volume, Watershed, Fast marching, and Flood Filling.
- Writing a coordinate system to model files.
- Adding OBJ, data dialog, and options for model nodes and for volume loading.
- Speeding up volume loading by reading scalar range from the header file.
- Loading of RGB volumes.
- Providing three views.
- Having a hierarchy view.
- Organizing data in folders or patient-subject trees.
- Having all nodes view (Robin, n.d.).
- Visualising of volumetric voxel images, polygonal meshes, and volume renderings.
- Manual editing.
- Registering of data by using rigid and non-rigid algorithms.
- Having automatic image segmentation (Pieper et al., 2006).

- Getting the CT scan of the spine from 3D slicer 4.8.1

By this software, your NRRD file is converted to the DICOM file format and then it is imported to the MIMICS software to making the geometry of vertebrae and the other parts (Picture 7-6, 7-7).

Pic 7-6: The Import of L3/ L4 CT scan data to 3d slicer for this thesis

Pic 7-7: CT scan data of this thesis in 3D slicer

7.1.1.1.3 Step three (MIMICS research 19.0)

Mimics is commercially available as part of the Materialise Mimics Innovation Suite, it consists of Mimics and 3-matic (a design and meshing software for anatomical data) (Picture 7-8). Mimics is an image processing software for 3D design and modeling from stacks of 2D image data such as Computed Tomography (CT), Magnetic Resonance Imaging (MRI) (Gelaude, Vander Sloten, & Lauwers, 2008; Halterman, n.d.), Micro CT, Confocal Microscopy, X-ray, and Ultrasound through image segmentation (Materialize, n.d.) (Picture 7-9a).

Materialise Mimics was developed by Materialise NV, a Belgian company for medical, dental, additive manufacturing industries, biomedical professionals (Halterman, n.d.), and a variety of engineering applications (Gelaude et al., 2008) (Picture 7-9b, 7-10, 7-11).

Pic 7-8: logo of Mimics software

Pic 7-9: (a) Mimics project example, (b) Spinal vertebrae displaying various mesh options. Top: normal high-density STL; Middle: surface mesh for FEA; Bottom: volume mesh including gray value-based material assignment

Pic 7-10: Image import from CT (DICOM), Image segmentation, 3D model

Pic 7-11: The process of changing CT scan data to volume in Mimics

- Advantages

- Remaining extremely intuitive and easy to learn software.
- Providing very powerful features and transforming into a 3D model easily with the ability of real-time rotation, pan, zoom, and transparency functions.
- Creating 3D models from imaging data easily and quickly.
- Measuring in 2D and 3D.
- Exporting 3D models in STL format.
- Exporting 3D models to 3-matic to optimize the mesh for FEA or CFD or to 3rd party software, like statistical, CAD, or FEA packages.
- Importing DICOM, JPEG, TIFF, BMP, or Raw image data (Materialize, n.d.).
- Representing the 3D files in the STL format (Picture 7-12).

Pic 7-12: The Mimics interface of working

- Making the human lumbar spine geometry in MIMICS

It happens by opening a DICOM file of human spine by its properties in MIMICS (Picture 7-13).

Pic 7-13: The properties of CT scan data in DICOM file in Mimics of this thesis

Verifying the orientation of CT scan (Picture 7-14).

Pic 7-14: The orientation of L3/ L4 in Mimics for this thesis

Importing file to MIMICS (Picture 7-15).

Pic 7-15: The Mimics file of this thesis

Making a mask for each vertebra and calculating them (Picture 7-16).

Pic 7-16: Calculation of L3/L4 for this thesis

Editing each mask slice by slice to receive the whole geometry of vertebrae (Picture 7-17, 7-18).

Pic 7-17: Making the Vertebrae in Mimics for this thesis

Pic 7-18: The final vertebrae shape in Mimics

Dividing each vertebra to cortical (around 3-5 mm around the exterior part of the vertebrae) and trabecular parts (just in exterior part) by editing slice by slice and calculating them (Picture 7-19).

Pic 7-19: The division of vertebrae in Mimics for this thesis

Making the annulus (in four layers) and nucleus (around 40% of the intervertebral disc) manually slice by slice and calculating them (Picture 7-20).

Pic 7-20: The annulus and nucleus shape for this thesis

Making cartilage end plates (the layer upper of L4 and below the L3 around 5 mm) and facet joints (the gap between two vertebrae around 8 mm) geometries manually slice by slice and calculating them (Picture 7-21).

Pic 7-21: The shape of endplate and facet joints for this thesis

Finally getting the vertebrae with two parts of cortical and trabecular, annular and nucleus parts of the intervertebral disc, end plates, and facet joints geometries (Picture 7-22) based on the CT scan.

Pic 7-22: The final shape of two-motion segment consists of vertebrae, annuals, nucleus, end plates, and facet joints

Then the geometries are prepared for the next step of making geometry by adding the others tissues such as annuals fibers and ligaments of the spine in the other software.

Making the volumes and meshing them are needed before adding the other geometries, so using the 3-Matic software is useful and proper here.

7.1.1.1.4 Step Four (3-matic (version 11.0))

It is intelligent software for data preparation, STL editor for 3D printing, additive manufacturing equipped and having a

Pic 7-23: The Logo of 3-Matic

customizable and intuitive user interface (“Materialise 3-Matic,” n.d.) (Picture 7-23). 3-matic is used by many industries such as interior design, concept cars, wind tunnel models, aerospace (Lightweight parts), medical (Porous structures), consumer goods, orthopedics and footwear (“3-Matic,” n.d.) (Picture 7-24).

Pic 7-24: The way of modelling with some examples

- Advantages

- Guiding efficiently through every step of your manufacturing and 3D printing (“Materialise 3-Matic,” n.d.).
- Optimizing and modifying designs at the STL level (“Materialise 3-Matic,” n.d.) with slice-based technology (“The Complete Software Suite for Professional 3D Printing,” n.d.) and split sets of triangles into surfaces (“3-Matic,” n.d.).
- Cleaning the raw data for simulation or converting a framework to CAD again (“3-Matic,” n.d.; “Materialise 3-Matic,” n.d.; “The Complete Software Suite for Professional 3D Printing,” n.d.).
- Enriching design by creating 3D textures (“Materialise 3-Matic,” n.d.; “The Complete Software Suite for Professional 3D Printing,” n.d.), perforations, patterns, lightweight models without compromising resistance (“Materialise 3-Matic,” n.d.) with a lattice structure (“The Complete Software Suite for Professional 3D Printing,” n.d.), and conformal structures for 3D printing (“3-Matic,” n.d.; “Materialise 3-Matic,” n.d.).
- Preparing data for a fast and efficient FE analysis (“Materialise 3-Matic,” n.d.) to compare easily (“3-Matic,” n.d.).

- Designing or repairing poorly scanned components by correcting problems (holes, edges, triangles, and other defects).
- Creating different values of flexibility and rigidity in the same piece (“Materialise 3-Matic,” n.d.).
- Meshing and remeshing the objects with new and advanced remapping algorithms to improve CAD formats (“3-Matic,” n.d.; “Materialise 3-Matic,” n.d.).
- Importing and exporting lattice structures to FEA (“Materialise 3-Matic,” n.d.; “The Complete Software Suite for Professional 3D Printing,” n.d.).
- Adapting the thickness of the beams to the lattice structures in 3D based on FEA.
- Checking the outlines of the triangles and graphs to highlight the problem areas.
- Having a User-friendly interface and getting better products to the market faster (“3-Matic,” n.d.).
- Importing almost all file formats and information with their native colors and original data (“Materialise 3-Matic,” n.d.).
- Adding logos, serial numbers, hollow parts, textures (“Materialise 3-Matic,” n.d.), create lattice structures, perform Boolean operations and advanced cuts (“3-Matic,” n.d.; “Materialise 3-Matic,” n.d.).
- Having the possibility of duplicating parts, improving orientation, generating support for metal parts and stereolithography, laser sintering, etc (“Materialise 3-Matic,” n.d.).
- Exporting to CAD (feature recognition can be applied) or FEA (“3-Matic,” n.d.) and converting your mesh back to CAD (“The Complete Software Suite for Professional 3D Printing,” n.d.).
- Having the flexibility to create shapes, which are impossible or too costly to produce with traditional technologies, and enhancing your design with aesthetic or functional textures, perforations, and patterns.
- Creating STL data from native and non-native CAD data such as IGES, VDA, VRML, Catia V4, V5, V6, UG, Pro-E, and STEP file formats.
- Optimizing the surface mesh through the different manual and automatic remesh operations by controlling the quality and size of the part’s surface triangulation.
- Offering a volume meshing functionality such as Tet4 and Tet10 volume meshes, and can be exported to Fluent, Ansys, Abaqus, Comsol, Patran, and Nastran software.
- Creating shapes which are impossible or too costly to produce with traditional technologies for 3D printing (“The Complete Software Suite for Professional 3D Printing,” n.d.).
- Smoothing, cleaning up and redesigning the raw output after topology optimization on STL level.

- Reducing the design cycles by going from scanned or CAD data to FEA or CFD application models ("3-Matic," n.d.) and replacing one part of the geometry by another geometry easily which are designed in 3-matic, or imported from a CAD package.
- Reducing the required post-processing time from days to hours.

- Materialise 3-Matic modeling of the human lumbar spine

Making copy and paste of masks from the 3D objects part of Mimics in the 3-magic software in the object tree parts, then it was smoothed (Picture 7-25).

Pic 7-25: The two motion segment of this thesis after smoothing in 3-Matic

Creating volume meshing of 10 tetrahedral and making uniform remesh of 1.5, and Filling with triangle edges (Picture 7-26, 7-27, 7-28, 7-29).

Pic 7-26: The endplate and facet joints of this thesis after meshing

Pic 7-27: The two motion segment of this thesis after getting mesh

Pic 7-28: The Annuals and nucleus of this thesis after meshing

The Geometries become ready for going to analyze, but before that, adding the annuals fibers and editing the geometry are needed. Therefore, we have to import the geometry in STL format to RHINO software for making them.

Pic 7-29: The views of two-motion segment of this thesis after meshing

7.1.1.1.5 Step Five (Rhinceros 5)

Rhinceros (Rhino or Rhino3D), is a 3D CAD modeling and computer-aided design software that are so accurate for modeling your designs and getting it ready for rendering, animation, drafting, engineering, analyzing (“Rhino 3D software: 3D Modeling Software for Architecture and Industrial Design,” n.d.; “Rhino software modeling tools for transforming 3D design into reality,” n.d.), computer-aided manufacturing (CAM), rapid prototyping, 3D printing, reverse engineering, automotive design, watercraft design, product design, multimedia, and graphic design (Novedge, n.d.) for architectural trades, industrial design, jewelry design (“Rhino 3D software: 3D Modeling Software for Architecture and Industrial Design,” n.d.; “Rhino software modeling tools for transforming 3D design into reality,” n.d.) (Picture 7-30).

Pic 7-30: The Logo of Rhinceros

Pic 7-31: The interface of Rhinceros

Rhino is a free-form NURBS surface modeler. With Rhino, you can create, edit, analyze, and translate NURBS curves, surfaces, and solids in Windows or Mac environments. There are no limitations on complexity, degree, or size of the model you design (RHINO SOFTWARE Modeling tools for transforming the 3D design into reality) (Picture 7-31).

Rhinoceros (Rhino, or Rhino3D) is developed by Robert McNeel & Associates (“Rhino 3D software: 3D Modeling Software for Architecture and Industrial Design,” n.d.) in 1980. Rhinoceros geometry is based on the NURBS mathematical model, which focuses on producing a mathematically precise representation of curves and freeform surfaces in computer graphics (as opposed to polygon mesh-based applications) (Novedge, n.d.). Rhinoceros received a Gold Customer Satisfaction Award from CNC Cookbook 2017 CAD Survey (Warfield, 2017).

- Advantages

- Creating a three-dimensional surface by the most accurate way of manipulating curves (NURBS) and adaptive mesh.
- Modifying a part designed using NURBS in a polygonal mesh.
- Performing Boolean operations (addition and subtraction of volumes) to create closed volumes which are suitable for 3D printing.
- Attaching two parts of an object in order to make a single object.
- Showing the edges of the model to correct the model and making solid and closed objects for 3D printing.
- Exporting your Rhino 3D modeling into many formats such as 3DS (3D studio), DXF (AutoCAD), SKP (Sketch up) and IGES (“Rhino 3D software: 3D Modeling Software for Architecture and Industrial Design,” n.d.) to due to its 3DM.
- Making changes in rendering and materials, then capturing the viewport easier and faster (Picture 7-32).

Pic 7-32: 3D models examples in RHINO

- Making 2D view and result faster, better, cleaner, and more customizable (Picture 7-33).

Pic 7-33: The 2D view of material in RHINO

- Refining the Geometry by disparate features into single commands (“What is new in Rhino 6 for windows?,” n.d.) (Picture 7-34).

Pic 7-34: Refine the mesh of geometry in RHINO

- Having a lot of plugins and an open set of development tools to make better documentation (“What is new in Rhino 6 for windows?,” n.d.) and facilitating 3D printing in specific fields like rendering, animation, architecture, marine, jewelry, engineering, prototyping, etc. (“Rhinoceros Resources,” n.d.) (Picture 7-35).

Pic 7-35: The plugins types for Rhino

- Having a free-form surface modeler that utilizes the NURBS mathematical model (“Rhinoceros Resources,” n.d.).
- The Rhinoceros file format is (.3DM) and it supports (without the use of external plugins): DWG, DXF (AutoCAD 200x, 14, 13, and 12), IGES, STEP, SolidWorks, SAT (ACIS, export only), Microstation DGN, Direct X (X file format), FBX X_T (Parasolid, export only), .3ds, LWO, STL, SLC, OBJ, AI, RIB, POV, UDO, VRML, CSV (export properties and hydrostatics), BMP, TGA, uncompressed TIFF, VDA, GHS, GTS, KML, PLY, and SketchUp file format (“Rhinoceros index of import/export file types,” n.d.).
- The following CAD file formats are supported with the use of external plugins: 3DPDF, ACIS, CATIA V4, CATIA V5, CATIA V6, GCR, Inventor, JT, Parasolid, PLMXML, Creo Parametric, Solid Edge, and Siemens NX (Thuilier, n.d.).

- Rhinoceros for the human lumbar spine

At first, importing the STL file format to the software, and then editing the geometry (Picture 7-36, 7-37).

Pic 7-36: The shape of annulus and nucleus of this thesis in RHINO

Pic 7-37: The two-motion segment of this Thesis in RHINO

Making the annual fibers with solid geometries in two layers with the shape of truss “X” shapes with the ± 10 degrees from each other in the vertical axis (Picture 7-38).

Pic 7-38: Modelling of annuals fibers of this thesis in RHINO

Then, there is a need for exporting the fibers in 3-Matic software to get its mesh again by an adaptive remesh icon (Picture 7-39, 7-40).

Pic 7-39: Meshing the annuals fibers of this thesis in 3-Matic

Pic 7-40: Meshing the annuals fibers of this thesis in 3-Matic

Then exporting the STL file format of them to Rhino again to make the whole geometry for analyzing in ANSYS (Picture 7-41).

Pic 7-41: The shape of annuals fibers of this thesis after meshing in RHINO

7.1.1.1.6 Step Six (ANSYS (workbench 19.0- Space claim part)

ANSYS is an American public company in Pennsylvania for developing and marketing engineering simulation software (H.-H. Lee, 2017) (Picture 7-42). ANSYS software is for designing products and semiconductors to test their durability, temperature distribution, fluid movements, and electromagnetic properties by making simulated computer models of structures, electronics, or machine components without building test products or conducting crash tests. ANSYS also develops software for data management and backup, academic research and teaching (H.-H. Lee, 2017) (“What is ANSYS,” n.d.) in virtually (“Ansys engineering software suites,” n.d.; “What is ANSYS,” n.d.) environment before manufacturing prototypes of products to determine and improve weak points, computer life and foresee probable problems by 3D simulations. This power comes from ANSYS solver algorithms with years of experience (“What is ANSYS,” n.d.) (Picture 7-43).

Pic 7-42: the logo of ANSYS

Pic 7-43: some examples of model in ANSYS

ANSYS was founded in 1970 by John Swanson when the engineers performed FEA by hand (H.-H. Lee, 2017). In the 2000s, Ansys made numerous gain of other engineering design companies, acquiring additional technology for fluid dynamics, electronics design, and other physics analysis such as: ICEM CFD Engineering software for mesh simulations 2000, Cadoc (computer-aided design company) in 2001, CFX (developed fluid dynamics simulation software) in 2003, Century Dynamics (a developer of hydrodynamics simulation tools) and Harvard Thermal Inc. (focused on simulating the cooling time and temperature of electronics) in 2005, Fluent Inc. with its fluid dynamics tools (the second-largest fluid dynamics company) in 2006, Ansoft Corporation (an electronics design provider) in 2008, Apache Design Solutions (focused on semiconductor simulation software) in 2011, Esterel Technologies (focused on simulating interactions between software and hardware) in 2012, Ansys (a cloud-based software company for engineering composites) in 2013, SpaceClaim (a 3-D modeling

company) in 2014, Gear Design Solutions and the systems analysis company Delcross Technologies in 2015, KPIT (an automotive design company) in 2016, 3DSIM (a 3D printing company) in 2017, and OPTIS (an optical simulation company) in 2018.

This ANSYS software contains some parts such as:

- Material book: representing material.
- Space claim: importing the geometry and adding the new other parts such as ligaments.
- Mechanical: getting extra material and properties for each part, connecting each part which is related to each other, meshing the adding parts, loading the bars, representing solutions, and getting the analyze data and its diagrams to compare with experimental data.

- Advantages

- Delivering more power and better results than other CAE tools due to its engineered scalability, a multiphysics foundation, and adaptive architecture.
- Solving most complex structural engineering projects.
- Providing the ability to automate, customize simulations, and parameterize for many design scenarios (“Ansys engineering software suites,” n.d.).
- Connecting easily to other physics tools for better realism and predicting performance and behavior of even the most complex projects (“Ansys engineering software suites,” n.d.; “What is ANSYS,” n.d.).
- Modeling nonlinearities (“Introduction to ANSYS Mechanical (Workbench),” n.d.).
- Simulating every structural aspect of a product to assess the strength, vibration, motion and thermal response characteristics of the system (“Mechanical Simulation Software,” n.d.).
- Visualizing the results of your simulation, showing you how best to modify your design and optimizing your product.
- ANSYS has a different type such as ANSYS Mechanical, ANSYS Structural, ANSYS Professional, ANSYS Design Space, and ANSYS Rigid Body Dynamics (ANSYS Workbench) (“Mechanical Simulation Software,” n.d.).
- Ensuring that your product will work as advertised (“Realize Your Product Promise,” n.d.).
- ANSYS can carry out advanced engineering analyses quickly (“What is ANSYS,” n.d.), safely and practically by its variety of contact algorithms, time-based loading features, and nonlinear material models (“What is ANSYS,” n.d.).
- Getting verification and improvement of the product in the virtual environment (“What is ANSYS,” n.d.) (Picture 7-44).

Pic 7-44: The types of ANSYS software such as Structures, Fluids, Electromagnetics, Semiconductors, Embedded Software, Systems, Partner Network, Optical, 3D

- The human lumbar spine in space claim

Importing the whole Geometry for adding the new other parts such as ligaments. Ligaments are a node to node beams with the profile of filled circle which is based on the area of real ligaments in the human spine (Picture 7-45, 7-46).

Pic 7-45: Adding the ligaments to the model of this thesis in Space claim parts of ANSYS

Pic 7-46: Adding the ligaments to the model of this thesis in Space claim parts of ANSYS

7.1.1.1.7 Step seven (ANSYS (workbench- Mechanical part))

According to this step, the model which is gained from the previous part is going to the mechanical part of ANSYS to get extra orders.

- Human lumbar spine in the mechanical part of ANSYS

Getting extra material properties for each part, connecting each part which is related to each other, and meshing the adding parts in space claim (Picture 7-47).

Pic 7-47: Meshing the ligaments in the Mechanical part of ANSYS

7.1.1.2 Representing the material of human lumbar spine in a material book

Representing material according to the mechanical behavior of each part which is explained in the tables (Table 7-1, 7-2) based on chapter five data (Picture 7-48).

Table 7-1: The whole Material properties of each part of two motion segments

Component	Element			Material properties			
	Type	Form	Area of contact (mm ²)	E (MPa)	V	Formulation	
Cortical bone	10 node tetrahedral (solid)	solid	CT scan data	12000	0.3	Linear	isotropic
Trabecular Bone	10 node tetrahedral (solid)	solid	CT scan data	140	0.2	Linear	isotropic
Cartilage endplate	10 node tetrahedral Hyper elastic	Mooney-Rivlin*	CT scan data	10.4	0.4	Linear	isotropic
Nucleus	10 node tetrahedral Hyper elastic	Mooney-Rivlin*	40% of Annuals matrix	1	0.49	Linear	isotropic
Annuals matrix (reinforced by collagen annuals fibers)	10 node tetrahedral Hyper elastic	Mooney-Rivlin*	CT scan data	4.2	0.45	Linear	isotropic
Annuals fiber 1	10 node tetrahedral Hyper elastic Truss	Mooney-Rivlin*	-	350	0.3	Linear	isotropic
Annuals fiber 2	10 node tetrahedral Hyper elastic Truss	Mooney-Rivlin*	-	455	0.3	Linear	isotropic
Facet Joints	10 node tetrahedral Hyperelastic	Mooney-Rivlin*	CT scan data	12	0.4	Linear	isotropic
ALL***	Beam – Tension only	Neo-Hookean**	5*12.56	10	0.3	Nonlinear	isotropic
PLL***	Beam– Tension only	Neo-Hookean**	5*4.15	20	0.3	Nonlinear	isotropic
LF***	Beam– Tension only	Neo-Hookean**	4*15.20	10	0.3	Nonlinear	isotropic
TL***	Beam– Tension only	Neo-Hookean**	2*4*1.15	20	0.3	Nonlinear	isotropic
CL***	Beam– Tension only	Neo-Hookean**	2*10*4.2	20	0.3	Nonlinear	isotropic
ISL***	Beam– Tension only	Neo-Hookean**	1*7	10	0.3	Nonlinear	isotropic
SSL***	Beam– Tension only	Neo-Hookean**	6*3.14	10	0.3	Nonlinear	isotropic
* Because of having the role of fluid ** Due to being not compressed material *** Anterior longitudinal ligament (ALL), Posterior longitudinal ligament (PLL), Flaval ligament (FL), Facet capsular ligament (CL), Intertransverse ligament (ITL), Interspinous ligament (ISL), and Supraspinous ligament (SSL)							

Table 7-2: The whole Geometry properties of each part of two motion segments in ANSYS

Component	Properties		
	Node	Element	Area
Cortical bone L3	158105	91872	0.098*0.096*0.053 m
Cortical bone L4	99104	56416	0.089*0.095*0.048 m
Trabecular Bone L3	42601	24875	0.048*0.038*0.027 m

Trabecular Bone L4	44298	25772	0.053*0.039*0.03 m
Cartilage endplate L3	15517	7570	0.05*0.041*0.004 m
Cartilage endplate L4	11347	5518	0.05*0.04*0.006 m
Nucleus	7588	4223	0.033*0.022*0.007 m
Annular matrix	59867	31820	0.054*0.042*0.013-0.007 m
Annular fiber 1	121072	46783	0.037*0.027*0.007 m
Annular fiber 2	257823	116634	0.042*0.031*0.008 m
Facet joints 1	3073	1662	0.015*0.013*0.017 m
Facet joints 2	2879	1526	0.013*0.016*0.017 m
ALL	5*17	5*8	5*12.56 (L:0.014-0.018)
PLL	5*17	5*8	5*4.15 (L:0.022-0.024)
LF	4*17	5*8	4*15.20 (L:0.022-0.025)
TL	8*17	8*8	2*4*1.15 (L:0.024-0.029)
CL	20*17	20*8	2*10*4.2 (L:0.0007-0.0055)
ISL	6*17	6*8	1*7 (L:0.0015-0.012)
SSL	1*17	1*8	6*3.14 (L:0.018)

Pic 7-48: The Material book interface of this thesis in ANSYS

7.1.1.3 Loading the bars of the human lumbar spine

This part contains how to load the bars and get its analyzed data from ANSYS. This model is under loads of both pure compression and pure moments. Pure compression is because of local muscle forces and upper body weight on these two motion segments and pure moments is because of the directional movements (Dreischarf et al., 2014; Xu et al., 2017) (Table 7-3).

Table 7-3: The properties of each load for each type of loading bars for this thesis

Body position	Compressive force (N)	Moment (Nm)	References
Flexion	1175	7.5	Rohlmann et al.(2009)
Extension	500	7.5	Rohlmann et al.(2009)
Lateral bending	700	7.8	Dreischarf et al.(2012)
Axial Rotation	720	5.5	Dreischarf et al.(2011)

For loading these bars, the bottom of the L4 vertebrae was fixed and the loads were applied on the top parts of L3 vertebrae (Picture 7-49).

Pic 7-49: Fixing the bottom part of L4

Loading bars happened in four different directions:

- Flexion (the combination of compression on top of L3 and the load of the pure moment of flexion on the half part of L3) (Picture 7-50).

Pic 7-50: The combination of compression on top of L3 and the Load of pure moment of flexion on the half part of L3

- Extension (the combination of compression on top of L3 and the load of the pure moment of extension on the half part of L3) (Picture 7-51).

Pic 7-51: The combination of compression on top of L3 and the Load of pure moment of Extension on the half part of L3

- Lateral bending (the combination of compression on top of L3 and the load of the pure moment of lateral bending on the half part of L3) (Picture 7-52).

Pic 7-52: The combination of compression on top of L3 and the Load of pure moment of lateral bending on the half part of L3

- Rotation (the combination of compression on top of L3 and the load of the pure moment of rotation on the whole of L3) (Picture 7-53).

Pic 7-53: The combination of compression on top of L3 and the Load of pure moment of rotation on the whole of L3

7.1.2 Processing step

This step contains representing the solution based on the goal of the study.

7.1.2.1 Representing solutions in the mechanical part of ANSYS

At this step, the solutions were chosen based on the things which have to validate and analyze in the next step of the FEM. So, the solutions which are used to get data for a further step of validating and analyzing step were chosen in the mechanical part of ANSYS.

This solution contains (picture 7-54):

- Equivalent (von-Mises) stress for whole geometry to measure the whole geometry stress.
- Equivalent Elastic Strain for whole geometry to measure the whole geometry strain.
- Equivalent (von-Mises) Stress for the nucleus part to measure the intradiscal pressure (IDP).
- Total deformation to get the deformation graph and motion of the whole geometry.

- The Differences of the angle between two vertebrae after and before of the loading bars by choosing the remote point with this command (Measure_Pilot=_npilot) and putting the command (Formula 7-1) for that remote point to get its data (ROM).

$$\text{my_rotx} = \text{ROTX}(\text{Measure_Pilot}) * 57.29577951 \quad (7-1)$$

$$\text{my_roty} = \text{ROTY}(\text{Measure_Pilot}) * 57.29577951$$

$$\text{my_rotx} = \text{ROTZ}(\text{Measure_Pilot}) * 57.29577951$$

Pic 7-54: The solutions for the Human lumbar spine

7.1.3 Post-processing step

At this step, the FE model of human lumbar spine has to analyze about their sensitivity and has to compare its verified data with the experimental vivo and vitro data, which are gathered in the Mechanical part.

7.1.3.1 Verified data

All data that are gained from validated resources which are explained in chapter five and the Table 7-1 and Table 7-2.

7.1.3.2 Sensitivity analyzing

For analyzing the two motion segments of the human lumbar spine, the new FE of human lumbar spine L3/ L4 was scaled. By these scaling, the features of two motion segments in normal size were constant, but the structure and its loads were scaled to keep the equivalent stress equal for them. So the two motion segments geometries were scaled 10 or 100 times by keeping the equivalent stress, meshing, geometry shapes, and materials equal with normal size, but the compression loads get 10^2 and 10^4 times and the moments get 10^3 and 10^6 times greater respectively according to Formulas 7-2 and 7-3.

7.1.3.2.1 The 10 times scaled greater of the human lumbar spine (L3/L4)

The two motion segments were scaled 10 times to analyze if this structure will be worked in this size (Picture 7-55).

Pic 7-55: 10 times greater of the FE of human lumbar spine

According to the two motion segments, this 10 times structure has the scale of $100 \times 100 \times 110 \text{ cm}^3$ with the equal mesh numbering of nodes and elements in its normal structures. However, the amount of compression load and moment were changed based on Formula 7-2, and the number of loads, which were applied to this motion segments, were according to Table 7-4.

$$\text{Stress (Pa)} = \sigma = F \text{ (N)} / A \text{ (m}^2\text{)} \quad (7-2)$$

$$\text{If stress} = \text{constant}, m_1 = m * 10 \rightarrow F_1 = F * 10^2$$

$$\text{Moment (Nm)} = F \text{ (N)} * d \text{ (m)} \rightarrow \text{stress} = (\text{moment}/d) / \text{m}^2 = \text{moment} / (d * \text{m}^2)$$

$$\text{If stress} = \text{constant}, m_1 = m * 10, d_1 = d * 10 \rightarrow \text{moment}_1 = \text{moment} * 10^3$$

Table 7-4: The loads for the structure with 10 times greater

Body position	Compressive force (N)	Moment (Nm)
Flexion	$1175 * 10^2$	$7.5 * 10^3$
Extension	$500 * 10^2$	$7.5 * 10^3$
Lateral Bending	$700 * 10^2$	$7.8 * 10^3$
Axial Rotation	$720 * 10^2$	$5.5 * 10^3$

7.1.3.2.2 The 100 times scaled greater of the human lumbar spine (L3/L4)

The two motion segments were scaled 100 times too to analyze if this structure will be worked for this size (Picture 7-56).

Pic 7-56: 10 times greater of the FE of human lumbar spine

According to the two motion segments, this 100 times structure has the scale of $10*10*11\text{ m}^3$ with the equal mesh numbering of nodes and elements in its normal structures, but the amount of compression load and moment were changed based on Formula 7-3. Therefore, the loads, which were applied to these motion segments, were according to Table 7-5.

$$\text{Stress (Pa)} = \sigma = F \text{ (N)} / A \text{ (m}^2\text{)} \quad (7-3)$$

$$\text{If stress=constant, } m_2 = m*100 \longrightarrow F_2 = F*10^4$$

$$\text{Moment (Nm)} = F \text{ (N)} * d \text{ (m)} \longrightarrow \text{stress} = (\text{moment}/d)/m^2 = \text{moment}/(d*m^2)$$

$$\text{If stress=constant, } m_2 = m*100, d_2 = d*100 \longrightarrow \text{moment}_2 = \text{moment}*10^6$$

Table 7-5: The loads for the structure with 100 times greater

Body position	Compressive force (N)	Moment (Nm)
Flexion	$1175*10^4$	$7.5*10^6$
Extension	$500*10^4$	$7.5*10^6$
Lateral Bending	$700*10^4$	$7.8*10^6$
Axial Rotation	$720*10^4$	$5.5*10^6$

7.1.3.3 Data comparison

At this step, for comparing data IDP and ROM data are gathered to compare with the experimental vivo and vitro data, which are explained before.

- Group one vitro data by Ming Xu, James Yang, Isador H. Lieberman, and Ram Haddas, 2016 for both ROM and IDP.
- Group two vitro data by Ayturk and Puttlitz, 2001, Kiapour et al., 2012a, Little et al., 2008, Liu et al., 2011, Park et al., 2013, Schmidt et al., 2012, Shirazi-Adl, 1994b, Zander et al., 2009 for both ROM and IDP.
- Vivo data by Percy et al. 1984, Percy & Tibrewal 1984, and Percy 1985 for ROM, and vivo data by Wilke et al. 2001 for IDP (Dreischarf et al., 2014; Xu et al., 2017).

7.2 Preliminary conclusions

The FE model of the human lumbar spine (L3/L4) was made based on CT scan data as a base of geometry, material data, and loading bar from verified data from validated literature references by different software in seven steps (Picture 7-57 (next page)). Then the ROM, IDP, stress, and strain data of this FE of the human lumbar spine were gathered in combined loading modes of pure compression and pure moments in four directions of flexion, extension, lateral bending, and axial rotation to compare with the experimental vivo and vitro data and checking its sensitivity.

Afterward, for checking the sensitivity of the FE of the human lumbar spine, the FE model was scaled 10 and 100 times greater by keeping its equivalent stress, mesh, material, and the form of geometry equal to the previous one. But the pure compression becomes 10^2 , 10^4 and

the pure moment becomes 10^3 and 10^6 greater for these 10 times and 100 times scaled greater respectively due to keeping the equivalent stress equal with the normal size. The ROM, IDP, stress, and strain data of this scaled geometry were gathered to compare with normal size data too. These models by having these scales and these capacities of moment and compression bearing loads can be useful for making future joint structures and building structures by having the features of the human spine such as aesthetic form, complex form, stable joint, dynamic form, modular shape, and light form.

Pic
7-57:

(Step one) the Lumbar CT scan, (Step two) DICOM file in 3d slicer, (Step three) the 3D form in MIMICS, (step four) Making volume, smoothing, and meshing in 3-matic, (step five) making annular fibers in Rhino, (step six) adding ligaments in the space claim part of ANSYS, (step seven) loading the bars in the mechanical part of ANSYS software.

CHAPTER VIII: SIMPLIFYING THE FE OF THE HUMAN LUMBAR SPINE

8.1 Simplifying the FE of the human lumbar spine (L3/L4)

(In order to create an elemental structure for future modeling)

In view of the facts and data presented in the previous chapters, the spine joint is a good example of flexible and adaptable natural structures the same as the other human joints (the frame of the human body). The spine joint like the other joints has six key features:

- Harmonizing aesthetic form by harmonizing in response to environmental forces (“Types of Joints,” n.d.; Zakharchuk, 2012).
- Complexity, by simultaneously facilitating for functions of flexion, extension, abduction, adduction, rotation, and circumduction (Zakharchuk, 2012).
- Comfort form by ensuring stable in different positions.
- Dynamic form by facilitating balance through flexible movements (Salsabili et al., 2018; Zakharchuk, 2012).
- Light structural system.
- Modular form (Salsabili et al., 2018; Taghizadeh & Bastanfard, 2012).

The roles perform by the spine joint is very similar to the type of role require within the building and industrial design; therefore, it is a natural element that can inform the future of architecture and industrial design. On the other hand, the spine joint as the most important and biggest structure of the human body joints system consists of vertebrae, intervertebral discs, ligaments, and facet joints (Salsabili, 2014). Nowadays by recognizing the spine joints, it is possible to simulate (Salsabili, 2014) the sustainable systems by increased efficiency. This simulation happens by using the combination of biology, construction, mathematics, mechanics, and computational programs and software (Salsabili et al., 2018, 2016). The simulation of such structure, however, would require a simplification of both the structural and material complexity of the spine joint in accordance with the architectural and construction resources available. This, in turn, would also ease the process of modeling of such structures both for the bioengineering and construction of prosthetics as well as other structures.

There are many FE simulations for investigating the biomechanical behavior of the lumbar spine, the largest part of the whole spine which is responsible for the majority of weight bearing and flexible movement functions (“Anatomy and Function,” n.d.; Kuo et al., 2010). As it currently stands, FE modeling had the simplification of the material parts of two motion segments of human lumbar spine L3/L4 during four continues steps to understand what may occur by changing or omitting these material parts on two-motion segments response. Then, the simplifications of the structural parts of two motion segments of human lumbar spine L3/L4 during four continues steps were accrued too to know what may occur by changing the structure parts of two motion segments while maintaining its mechanical and anatomical properties of human lumbar spine to carry out the simple and basic structure by FE method based on the human lumbar spine which is the best joints in nature by having static and

dynamic behavior against the internal and external loads. In the end, the structure was scaled to check it for bigger size structure.

In the following, there are these levels by their subsets and their data, which are gained from them to analyze for getting a better structure. In each step, the changing added to the previous steps, not instead of them.

8.2 Changing the material part of two motion segments

The materials of the two motion segments of the human lumbar spine consist of the fourteen different materials with different Modulus of elasticity and Poisson's ratio. For making a simple structure, the fewer materials may be useful to make the structure easier and less expensive. Therefore, at this part, some materials, which can be omitted or changed, were chosen to test.

8.2.1 Step one (The two motion segments without annular fibers)

The annular fibers are worked just in extension, in this level we are going to omit it to know how much they work in different positions of flexion, extension, lateral bending, and axial rotation (Picture 8-1).

Pic 8-1: Omitting the annular fibers

8.2.2 Step two (Representing the material of annular to the nucleus)

To make the simple role for an intervertebral disc, the nucleus material was changed with annular material to analyze its effect (Picture 8-2).

Pic 8-2: Representing the material of annular to the nucleus

8.2.3 Step three (Representing the material of annulus to the endplates too)

The second work to simplify the intervertebral disc structure was to change the endplate material to the annulus material (Picture 8-3).

Pic 8-3: Representing the material of the annulus to the endplates

8.2.4 Step four (Omitting the trabecular part of vertebrae)

Making the structure based on the two motion segments needs an empty place in the middle of the vertebrae, so by omitting the trabecular part, there is the capacity of measuring its effects (Picture 8-4).

Pic 8-4: Omitting the trabecular part of vertebrae

8.3 Changing the structure of two motion segments

Until this part, the simplifications were in the material parts of the human lumbar spine. That simplification is useful to make any type of structures such as prosthesis or building structures and it may be useful for different field of studying such as biomechanics, industrial design, and building structure and it may be useful for both inside or outside of the body. Now we are going to simplify the structure to make the elemental structure, which is useful for further structure for industrial or building structures. The further steps were happened for the human lumbar spine structure in four continuous steps.

8.3.1 Step five (Simplify the main part of vertebrae)

The vertebrae have a little rippling surface, so by simplifying these surfaces, there is the capacity of measuring its effects. In this simplification, the shape and the scale of the main part of vertebrae didn't change (Picture 8-5).

Pic 8-5: The model of two motion segments structure by changing the main part of vertebrae

8.3.2 Step six (Simplify the posterior part of the vertebrae)

At this step, the posterior part of vertebrae was changed to the simplified structure by keeping the scale of the main part the same as the normal vertebrae. By changing the posterior parts, the TL, LF, PLL, SSL, and ISL ligaments were changed too. The ligaments in each side are according to ALL ligaments because of the new roles and the CL ligaments are the same as real one (Picture 8-26).

Pic 8-6: The model by changing the posterior part of vertebrae to make a simple structure

8.3.3 Step seven (Changing the main part of the vertebrae to the cylinder)

At these steps due to having the simple structure, the shape of the intervertebral disc is changed to the circle. Therefore, the vertebrae structures were changed to the cylinder to have the circle plan too and the posterior part became a little wider than before (Picture 8-7).

Pic 8-7: the model by changing the disc and the main part of the vertebrae to the circle plan

8.3.4 Step eight (The final shape with cylinder vertebrae and not modular)

At these steps, the shape of two motion segments was changed to be not modular for the structure, which is wanted to be in just to parts (Picture 8-8).

Pic 8-8: the new structure with the cylinder (main part of vertebrae) and not modular

8.4 Represent the new structures for future structures

(Based on the two motion segments of the human lumbar spine (L3/L4))

The new elemental and basic structure was made during eight steps of simplifications. Now, it is the time of checking this structure for making the joint and building structure by checking its sensitivity of the structure in greater size. Based on the previous part, it happens by

applying features of the new two motion segments in normal size by changing the scale of that structure and its loads to keep the equivalent stress equal for them. So the new geometries were scaled 10 or 100 times by keeping the equivalent stress, meshing, geometry shapes, and materials equal for them, but the compression loads get 10^2 and 10^4 times and the moments get 10^3 and 10^6 times greater respectively according to Formula 8-1, 8-2.

8.4.1 The new structure with 10 times scaled greater

(It is useful for making the type of joint in buildings)

The new two motion segments were scaled 10 times to analyze if this structure will be useful as the joints in building structures or not (Picture 8-9).

Pic 8-9: 10 times greater of the new structure

According to these new segments, this 10 times structure has the scale of $50 \times 50 \times 68 \text{ cm}^3$ with the equal mesh numbering of nodes and elements in its structures with the normal size. The amount of compression load and moment were changed based on Formula 8-1, and the number of loads, which were applied to these motion segments, was according to Table 8-1.

$$\text{Stress (Pa)} = \sigma = F \text{ (N)} / A \text{ (m}^2\text{)} \quad (8-1)$$

$$\text{If stress} = \text{constant}, m_1 = m * 10 \longrightarrow F_1 = F * 10^2$$

$$\text{Moment (Nm)} = F \text{ (N)} * d \text{ (m)} \longrightarrow \text{stress} = (\text{moment}/d)/m^2 = \text{moment}/(d * m^2)$$

$$\text{If stress} = \text{constant}, m_1 = m * 10, d_1 = d * 10 \longrightarrow \text{moment}_1 = \text{moment} * 10^3$$

Table 8-1: The loads for the structure with 10 times greater

Body position	Compressive force (N)	Moment (Nm)
Flexion	$1175 * 10^2$	$7.5 * 10^3$
Extension	$500 * 10^2$	$7.5 * 10^3$
Lateral Bending	$700 * 10^2$	$7.8 * 10^3$
Axial Rotation	$720 * 10^2$	$5.5 * 10^3$

8.4.2 The new structure with 100 times scaled greater

(It is useful for making the building)

The new two motion segments were scaled 100 times too to analyze if this structure will be useful as the structure of the building or not (Picture 8-10).

Pic 8-10: 100 times greater of the new

According to the new two motion segments, this 100 times structure has the scale of $5*5*6.8 \text{ m}^3$ with the equal mesh numbering of nodes and elements in its structures with the normal size, and the amount of compression load and moment were changed based on Formula 8- 2. So, the loads, which were applied to these motion segments, were according to Table 8- 2.

$$\text{Stress (Pa)} = \sigma = F \text{ (N)} / A \text{ (m}^2\text{)} \quad (8-2)$$

$$\text{If stress=constant, } m_2 = m * 100 \longrightarrow F_2 = F * 10^4$$

$$\text{Moment (Nm)} = F \text{ (N)} * d \text{ (m)} \longrightarrow \text{stress} = (\text{moment}/d)/m^2 = \text{moment}/(d*m^2)$$

$$\text{If stress=constant, } m_2 = m * 100, d_2 = d * 100 \longrightarrow \text{moment}_2 = \text{moment} * 10^6$$

Table 8-2: The loads for the structure with 100 times greater

Body position	Compressive force (N)	Moment (Nm)
Flexion	$1175*10^4$	$7.5*10^6$
Extension	$500*10^4$	$7.5*10^6$
Lateral Bending	$700*10^4$	$7.8*10^6$
Axial Rotation	$720*10^4$	$5.5*10^6$

8.5 Preliminary conclusions

After conducting the FE of human lumbar spine L3/L4 by software and scaling it 10 times and 100 times to check it for bigger scaled sizes and sensitivity, it is the time of simplification that structure by keeping its mechanical and anatomical behavior. At this chapter, we faced the steps of the simplification of that structure in eight continuous steps. The first four steps were in the material part and the second four steps were in the mechanical parts of the human lumbar spine L3/L4 to make the basic structure. For analyzing and validating the structure, we get the IDP, ROM, Stress curves, and strain curves of these steps from ANSYS software for comparing them with the experimental data for analyzing its internal and external response.

After making the new structure based on the human lumbar spine L3/L4, we were going to scale this new structure to check its capacity of that for bigger size structure for building joins

and building structure. These scaled structures are under a large amount of loading according to loads of human lumbar spine. So if this structure is validated by its data, could be a good example of a structure for tolerating the much loads in compression and moment.

The chart below shows the way and steps of simplification to receive a new structure for further study (Picture 8-11).

Levels	Steps	Works description	Output form
Material	Step one	The two motion segments without annals fibers
	Step two	Representing the material of annals to the nucleus
	Step three	Representing the material of annals to the endplates too
	Step four	Omitting the trabecular part of vertebrae
Structural	Step five	Simplify the main part of vertebrae
	Step six	Simplify the posterior part of the vertebrae
	Step seven	Changing the main part of the vertebrae to the cylinder
	Step eight	The final shape with cylinder vertebrae and not modular

Pic 8-11: The way and steps of material and

CHAPTER IX: RESULTS

9.1 Results parts

This research contains four main steps,

- Making the FE of human lumbar spine L3/L4 by FEM based on the human lumbar spine verified anatomical and mechanical data.
- Simplifying the FE element of the human lumbar spine by its material parts.
- Simplifying the FE element of the human lumbar spine by its structural parts.
- Making the new structure for building joints and building structure.

All these steps have to be checked their results to understand if these steps are validated or not.

9.2 The finite element of human lumbar spine L3/L4 results

According to the methods part and solution which were chosen, there are three parts of deformation distribution data, comparing data (IDP, ROM, stress, and strain), and sensitive data as a result for analyzing.

9.2.1 Deformation distribution data

- Flexion: the dispersion of loads of flexion on the two motion segments (Picture 9-1).

Pic 9-1: the dispersion of the loads of flexion on the two motion segments

- Extension: the dispersion of loads of extension on the two motion segments (Picture 9-2).

Pic 9-2: the dispersion of the loads of extension on the two motion segments

- Lateral bending: the dispersion of loads of lateral bending on the two motion segments (Picture 9-3).

Pic 9-3: the dispersion of loads of lateral bending on the two motion segments

- Rotation: the dispersion of loads of axial rotation on the two motion segments (Picture 9-4).

Pic 9-4: the dispersion of the loads of rotation on the two motion segments

9.2.2 Comparing data

For this step, some criteria were defined for validation of the FE model of two motion segments as a solution. These are the Range of motion (ROM) for external response and the intradiscal pressure (IDP) for an internal response, which are measured to compare with experimental vivo and vitro data (Diagram 9-1).

ROM is the differences between the angles of two vertebrae before and after of loading the bars in all four directions of flexion, extension, lateral bending, and axial rotation in degree.

IDP is the pressure in the midpoint of the nucleus of the intervertebral disc in all four directions of flexion, extension, lateral bending, and axial rotation in MPa.

Diag 9-1: The data for all four positions of flexion, extension, lateral bending, and axial rotation for normal size (a) Intradiscal pressure (IDP) in MPa, (b) Range of motion (ROM) in degree, (c) the strain curve in m/m, and (d) the stress curve in MPa

IDP data: the intervertebral pressure for L3/L4 are 1.01, 0.18, 0.51, and 0.51 MPa under its loadings for flexion, extension, lateral bendings, and axial rotation respectively.

ROM data: the range of motion for L3/L4 are 1.16, 0.11, 0.78, and 0.59 degrees under its loadings for flexion, extension, lateral bending, and axial rotation respectively.

Stress curves: the stress curves was made based on the data of the FE method. Maximum stress happened in inner annular fiber and the minimum happened in the outer annular fibers in all four position of loadings.

Strain curves: the strain curves were made based on the data of the FE method too. Maximum strain happened in inner annular fiber and the minimum happened in the outer annular fibers too in all four position of loadings.

9.2.3 Sensitivity data

9.2.3.1 10 times scaled data

The data which were gotten from the ANSYS are shown in Diagram 9-2.

Diag 9-2: The data for all four positions of flexion, extension, lateral bending, and axial rotation for 10 times scaled (a) Interdiscal pressure (IDP) in MPa, (b) Range of motion (ROM) in degree, (c) the strain curve in m/m, and (d) the stress curve in MPa.

IDP: the intervertebral pressure for L3/L4 in 10 times scaled greater are 1.10, 0.18, 0.55, and 0.541 MPa under its loadings for flexion, extension, lateral bending, and axial rotation respectively.

ROM: the range of motion for L3/L4 in 10 times scaled greater are 1.39, 0.2, 0.91, and 0.64 degrees under its loadings for flexion, extension, lateral bending, and axial rotation respectively.

Stress curves: the minimum stress in 10 times scaled greater happened in the outer annulars fiber and the maximum happened in the inner annulars fiber.

Strain curves: the minimum stress in 10 times scaled greater happened in the outer annulars fiber and the maximum happened in the inner annulars fiber.

9.2.3.2 100 times scaled data

The data which were gotten from the ANSYS are shown in Diagram 9-3.

Diag 9-3: The data for all four positions of flexion, extension, lateral bending, and axial rotation for 100 times scaled (a) Interdiscal pressure (IDP) in MPa, (b) Range of motion (ROM) in degree, (c) the stress curve in MPa, and (d) the strain curve in m/m.

IDP: the intervertebral pressure for L3/L4 in 100 times scaled greater are 1.10, 0.18, 0.55, and 0.542 MPa under its loadings for flexion, extension, lateral bending, and axial rotation respectively.

ROM: the range of motion for L3/L4 in 100 times scaled greater are 1.40, 0.2, 0.91, and 0.64 degrees under its loadings for flexion, extension, lateral bending, and axial rotation respectively.

Stress curves: the minimum stress in 100 times scaled greater happened in the outer annular fiber and the maximum happened in the inner annular fiber.

Strain curves: the minimum stress in 100 times scaled greater happened in the outer annular fiber and the maximum happened in the inner annular fiber.

9.3 Changing the material part of two motion segments data

This part contains four steps of material simplifications (MS) data.

9.3.1 Step one data (The two motion segments without annular fibers)

Diag 9-4: The data for all four positions of flexion, extension, lateral bending, and axial rotation for L3/L4 without annular fibers (a) Intradiscal pressure (IDP) in MPa, (b) Range of motion (ROM) in degree, (c) the strain curve in m/m, and (d) the stress curve in MPa.

By omitting the annuals fiber which is worked just in extension, we faced the IDP of 1.057, 0.187, 0.529, and 0.528 MPa and ROM of 1.20, 0.118, 0.78, and 0.6 degrees for flexion, extension, lateral bending, and axial rotation respectively. The stress and strain curves showed that the maximum stress of 77.3 MPa happened in a cortical part of L3 in the lateral bending direction and the maximum strain of 1.32 mm happened in annuals part in the flexion (Diagram 9-4).

9.3.2 Step two data (Representing the material of annuals to the nucleus)

Diag 9-5: The data for all four positions of flexion, extension, lateral bending, and axial rotation for L3/L4 without annuals fibers and with the same material for nucleus and annuals matrix (a) Intradiscal pressure (IDP) in MPa, (b) Range of motion (ROM) in degree, (c) the strain curve in m/m, and (d) the stress curve in MPa.

By changing the material of the nucleus by representing the material of annual to make the simple intervertebral disc, we faced the IDP of 1.55, 0.187, 0.77, and 0.76 MPa and ROM of 1.15, 0.12, 0.74, and 0.54 degree for flexion, extension, lateral bending, and axial rotation

respectively. The stress and strain curves showed that the maximum stress of 73.9 MPa happened in a cortical part of L3 in the lateral bending direction and the maximum strain of 1.23 mm happened in annular part in the flexion (Diagram 9-5).

9.3.3 Step three data (Representing the material of annulars to the endplates)

Diag 9-6: The data for all four positions of flexion, extension, lateral bending, and axial rotation for L3/L4 without annular fibers and with the same material for nucleus, annulars matrix, and endplates (a) Intradiscal pressure (IDP) in MPa, (b) Range of motion (ROM) in degree, (c) the strain curve in m/m, and (d) the stress curve in MPa.

By changing the material of the endplates the same as the previous step by representing the material of annulars to make the intervertebral disc with one material, we faced the IDP of 1.54, 0.267, 0.772, and 0.768 MPa and ROM of 1.27, 0.105, 0.83, and 0.6 degree for flexion, extension, lateral bending, and axial rotation respectively. The stress and strain curves showed

that the maximum stress of 81.8 MPa happened in a cortical part of L3 in the lateral bending direction and the maximum strain of 1.78 mm happened in annuals part in the flexion (Diagram 9-6).

9.3.4 Step four data (Omitting the trabecular part of vertebrae)

Diag 9-7: The data for all four positions of flexion, extension, lateral bending, and axial rotation for L3/L4 without annuals fibers, without trabecular part of vertebrae, and with the same material for nucleus, annuals matrix, and endplates (a) Interdiscal pressure (IDP) in MPa, (b) Range of motion (ROM) in degree, (c) the strain curve in m/m, and (d) the stress curve in MPa.

By omitting the trabecular part of vertebrae to make the light structure with one material, we faced the IDP of 1.495, 0.263, 0.747, and 0.741 MPa and ROM of 1.23, 0.105, 0.77, and 0.52 degree for flexion, extension, lateral bending, and axial rotation respectively. The stress and strain curves showed that the maximum stress of 83 MPa happened in a cortical part of L3 in the lateral bending direction and the maximum strain of 1.82 mm happened in annual part in the flexion (Diagram 9-7).

9.4 Changing the structure of two motion segments data

Until this part, the simplifications were in the material parts of the human lumbar spine. Now we are going to simplify the structure in structure to make the elemental structure which is useful for further structure for industrial or building structures. The further steps happened in the structure in four continuous steps.

9.4.1 Step five data (Simplify the main part of vertebrae)

Diag 9-8: The data for all four positions of flexion, extension, lateral bending, and axial rotation for L3/L4 without annulus fibers, without trabecular part of vertebrae, with the same material for nucleus, annulus matrix, and endplates, and changing the main part of vertebrae (a) Intradiscal pressure (IDP) in MPa, (b) Range of motion (ROM) in degree, (c) the strain curve in m/m, and (d) the stress curve in MPa.

By changing the main part of vertebrae to make the simple form, we faced the IDP of 0.817, 0.163, 0.543, and 0.42 MPa and ROM of 0.87, 0.104, 0.7, and 0.55 degree for flexion, extension, lateral bending, and axial rotation respectively. The stress and strain curves showed that the maximum stress of 251 MPa happened in a cortical part of L3 in the flexion direction and the maximum strain of 0.203 mm happened in intervertebral disc in the flexion (Diagram 9-8).

9.4.2 Step six data (Simplify the posterior part of the vertebrae)

Diag 9-9: The data for all four positions of flexion, extension, lateral bending, and axial rotation for L3/L4 without annulus fibers, without trabecular part of vertebrae, with the same material for nucleus, annulus matrix, and endplates, and changing the main and posterior part of vertebrae (a) Intradiscal pressure (IDP) in MPa, (b) Range of motion (ROM) in degree, (c) the strain curve in m/m, and (d) the stress curve in MPa.

By simplifying the posterior part of vertebrae to make the simple form, we faced the IDP of 1.02, 0.441, 0.552, and 0.487 MPa and ROM of 1.38, 0.61, 0.66, and 0.8 degrees for flexion, extension, lateral bending, and axial rotation respectively. The stress and strain curves showed that the maximum stress of 136 MPa happened in a cortical part of L3 in the flexion direction and the maximum strain of 0.265 mm happened in intervertebral disc in the flexion (Diagram 9-9).

9.4.3 Step seven data (Changing the main part of the vertebrae to the cylinder)

Diag 9-10: The data for all four positions of flexion, extension, lateral bending, and axial rotation for L3/L4 without annular fibers, without trabecular part of vertebrae, with the same material for nucleus, annular matrix, and endplates, and changing the main and posterior part of vertebrae to cylinder shape (a) Interdiscal pressure (IDP) in MPa, (b) Range of motion (ROM) in degree, (c) the strain curve in m/m, and (d) the stress curve in MPa.

By changing the main part of vertebrae to cylinder form, we faced the IDP of 1.02, 0.231, 0.533, and 0.441 MPa and ROM of 1.14, 0.258, 0.51, and 0.475 degrees for flexion, extension, lateral bending, and axial rotation respectively. The stress and strain curves showed that the maximum stress of 68.3 MPa happened in a cortical part of L3 in the extension direction and the maximum strain of 0.244 mm happened in intervertebral disc in the flexion (Diagram 9-10).

9.4.4 Step eight data (The shape with cylinder vertebrae and non-modular)

Diag 9-11: The data for all four positions of flexion, extension, lateral bending, and axial rotation for L3/L4 without annular fibers, without trabecular part of vertebrae, with the same material for nucleus, annular matrix, and endplates, and changing the main and posterior part of vertebrae to cylinder shape without modular form (a) Intradiscal pressure (IDP) in MPa, (b) Range of motion (ROM) in degree, (c) the strain curve in m/m, and (d) the stress curve in MPa.

By changing the vertebrae of the new motion segments to the not modular shape, we faced the IDP of 0.962, 0.247, 0.484, and 0.397 MPa and ROM of 0.882, 0.357, 0.43, and 0.41degree for flexion, extension, lateral bending, and axial rotation respectively. The stress and strain curves showed that the maximum stress of 109 MPa happened in a cortical part of L3 in the flexion direction and the maximum strain of 0.229 mm happened in intervertebral disc in the flexion (Diagram 9-11).

9.5 Representation of the new structures for future structures data

(Based on the two motion segments of the human lumbar spine (L3/L4))

The new elemental and basic structure was made during 8 steps of simplifications. Now, it is the time of checking this structure for making the joint and building structure. Based on the previous chapter (chapter 7), it happens by applying features of the new two motion segments in normal size with changing the scale of that structure and it's loads by keeping the equivalent stress equal for them. So the new geometries were scaled 10 or 100 times by keeping the equivalent stress, meshing, geometry shapes, and materials equal for them, and the compression loads get 10^2 and 10^4 times and the moments get 10^3 and 10^6 times greater respectively according to Formula 7- 1, 7- 2.

9.5.1 The new structure with 10 times scaled greater data

(It is useful for making the type of joint in buildings)

Diag 9-12: The data for all four positions of flexion, extension, lateral bending, and axial rotation for the new structure in 10 times scaled (a) Interdiscal pressure (IDP) in MPa, (b) Range of motion (ROM) in degree, (c) the strain curve in m/m, and (d) the stress curve in MPa.

IDP data: the intervertebral pressure for the new structure in 10 times scaled greater are 1.183, 0.312, 0.591, and 0.473 MPa under its loadings for flexion, extension, lateral bending, and axial rotation respectively.

ROM data: the range of motion for the new structure in 10 times scaled greater are 1.126, 0.47, 0.58, and 0.51 degrees under its loadings for flexion, extension, lateral bending, and axial rotation respectively.

Stress curves: the minimum stress in the new structure with 10 times scaled greater was happened in the vertebrae L4 in all directions and the maximum happened in the Vertebrae L3.

Strain curves: the minimum stress in the new structure with 10 times scaled greater was happened in the vertebrae L4 in all directions and the maximum happened in the disc in flexion and facet joints in all the other directions (Diagram 9-12).

9.5.2 The new structure with 100 times scaled greater data

(It is useful for making the building structure)

Diag 9-13: The data for all four positions of flexion, extension, lateral bending, and axial rotation for the new structure in 100 times scaled (a) Interdiscal pressure (IDP) in MPa, (b) Range of motion (ROM) in degree, (c) the stress curve in MPa, and (d) the strain curve in m/m.

IDP data: the intervertebral pressure for the new structure in 100 times scaled greater are 1.167, 0.312, 0.594, and 0.47 MPa under its loadings for flexion, extension, lateral bending, and axial rotation respectively.

ROM data: the range of motion for the new structure in 100 times scaled greater are 1.126, 0.47, 0.58, and 0.51 degrees under its loadings for flexion, extension, lateral bending, and axial rotation respectively.

Stress curves: the minimum stress in the new structure with 100 times scaled greater was happened in the vertebrae L4 in all directions and the maximum happened in the Vertebrae L3.

Strain curves: the minimum stress in the new structure with 100 times scaled greater was happened in the vertebrae L4 in all directions and the maximum happened in the disc in flexion and facet joints in all the other directions (Diagram 9-13).

CHAPTER X: DISCUSSIONS

10.1 Discussions part

This chapter systematically discusses all the results gained from all experiments carried out during this study. All analytical results are presented in the form of pictures and diagrams, and the key features are subsequently discussed. Further discussions are made over the gained results and it has been tried by the author to demonstrate the results in the form of diagrams to make them more understandable. This chapter consists of analyzed results gained from characterization sets along with the results of the modeling part. Validation of results obtained by comparing them with experimental vivo and vitro data to receive the goals of making the new structure and models for further buildings.

According to previous chapters, this modeling contains some levels which are described below.

10.2 The finite element of human lumbar spine L3/L4 Data

At the first level of modeling, this research contains an FE model of the L3/L4 by software (Diagram 8-1). Now we are going to show the comparison of its results by the experimental vivo and vitro data (Diagrams 10-1, 10-2) to validate it.

The model was under static loading condition of pure compression and pure moment for each direction and its data were checked for an internal response (IDP) and external response (ROM) and then getting their stress and strain curves for analyzing them for further researching. In the analysis of the data, the Von-Mises stress (the equivalent of stress) was calculated as its effects of the hydraulic behavior of the intervertebral disc and facets joints.

10.2.1 Comparing the FE of human lumbar spine data by experimental data

For validating these data, the simulation results of the FE models in this research were compared with the four groups of experimental data which are mentioned before such as:

- The “Previous study” by Ming Xu, James Yang, Isador H. Lieberman, and Ram Haddas, 2016 in both ROM and IDP data (Xu et al., 2017).
- The “Median and the range of max and min of these eight studies” by Ayturk and Puttlitz, 2011 (Ayturk & Puttlitz, 2011), Kiapour et al., 2012 (Kiapour, Ambati, Hoy, & Goel, 2012), Little et al., 2008 (Little, De Visser, Percy, & Adam, 2008), Liu et al., 2011 (Liu et al., 2011), Park et al., 2013 (W. M. Park, Kim, & Kim, 2013), Schmidt et al., 2012 (Schmidt, Galbusera, Rohlmann, Zander, & Wilke, 2012), Shirazi-Adl, 1994 (Shirazi-Adl, 1994), and Zander et al., 2009 (Zander, Rohlmann, & Bergmann, 2009) in both ROM and IDP data (Dreischarf et al., 2014).
- 3- The “vivo data” by Percy et al. 1984 (M. Percy, Portek, & Shepherd, 1984), Percy & Tibrewal 1984 (M J Percy & Tibrewal, 1984), and Percy 1985 (Mark J Percy, 1985) in ROM data and the “vivo data” by Wilke et al. 2001 in IDP data (Dreischarf et al., 2014; Wilke, Neef, Hinz, Seidel, & Claes, 2001; Xu et al., 2017).

The analyzed data of IDP, ROM, stress curve and strain curve for all four directions of flexion, extension, lateral bending, and axial rotation which were gained from the ANSYS software,

Diag 10-1: Intervertebral pressure (IDP) for two motion segments of this study in compare of the other studies (a) in flexion, (b) in extension, (c) in lateral bending, (d) in axial rotation.

is showed in diagram 8-1 and the Diagrams 10-1, 10-2 are the comparison of the results of this modeling with the previously validated data.

IDP data: there is a lack of vivo data for the L3/L4 motion segments, so we have to compare these data with L4/L5 motion segments data.

Flexion data: this data is near the minimum of the vitro data but it has 0.6 MPa differences with vivo data (Diagram 10-1a).

Extension data: this data is near the minimum of the vitro data too but it has 0.4 MPa differences with vivo data (Diagram 10-1b).

Lateral bending data: this data is more similar to the mean value of vitro data and has just 0.1 MPa differences with vivo data (Diagram 10-1c).

Axial rotation data: this data is more similar to the mean value of vitro data too and has just 0.1 MPa differences with vivo data (Diagram 10-1d).

Diag 10-2: Range of motion (ROM) for two motion segments of this study in compare of the other studies (a) in flexion, (b) in extension, (c) in lateral bending, (d) in axial rotation.

ROM data

Flexion data: this data has 3 degrees differences with vitro data and 7 degrees with vivo data. However, there is a big difference between vivo and vitro data in this group too and the vivo data is 7 degrees more than the vitro data (Diagram 10-2a).

Extension data: this data is more similar to mean of vivo data and it has 2.5 degrees differences with vitro data. But the vitro data is 2.5 degrees more than the vivo experimental data (Diagram 10-2b).

Lateral bending data: this data has 2 degrees differences with mean of vitro data and 3 degrees with a mean of vivo data. The mean vivo data and vitro data are similar and around 1-degree differences (Diagram 10-2c).

Axial rotation data: this data has just 0.5-degree differences with the mean of vivo and vitro experimental data. And the experimental data for vivo and vitro data are more similar to each other (Diagram 10-2d).

Stress curves: the stress curves were made based on the data of the FE method and were displayed in (Diagram 9-1c). Maximum stress happened in inner annuals fiber and the minimum happened in the outer annuals fibers in all four position of loadings.

Strain curves: the strain curves were made based on the data of the FE method too and was displayed in (Diagram 9-1d). The Maximum strain happened in inner annuals fiber and the minimum happened in the outer annuals fibers too in all four position of loadings.

10.2.2 Validating the FE of the human lumbar spine data

The FE model of the L3/L4 before simplification under both pure compression and the pure moment in four directions of flexion, extension, lateral bending, and axial rotation was validated with checking its internal response (IDP) and external response (ROM) by the experimental vivo and vitro data (Diagrams 10-1 and 10-2) (Dreischarf et al., 2014; Xu et al., 2017). For comparing the FE data with experimental data, at first, the experimental data were analyzed. All the experimental vitro data in two groups (Group one vitro data by Ming Xu, James Yang, Isador H. Lieberman, and Ram Haddas, 2016 and Group two vitro data by Ayturk and Puttlitz, 2001, Kiapour et al., 2012a, Little et al., 2008, Liu et al., 2011, Park et al., 2013, Schmidt et al., 2012, Shirazi-Adl, 1994b, Zander et al., 2009) have the similar data for both ROM and IDP data for human lumbar spine (L3/L4) in all four directions of flexion, extension, lateral bending, and axial rotation. The experimental vivo data for IDP by Wilke et al. 2001 for human lumbar spine L4/L5 due to the lack of data for human lumbar spine L3/L4 have a little difference with the media of IDP vitro data especially in the flexion and extension but these data are in the range of experimental vitro IDP data (Diagram 10-1). The experimental vivo data for ROM by Percy et al. 1984, Percy & Tibrewal 1984, and Percy 1985 for human lumbar spine L3/L4 are more similar to the experimental vitro ROM data except for the flexion data which has 8 degrees differences with the vitro ROM data and it is out of the range of vitro data too (Diagram 10-2) (Dreischarf et al., 2014; Xu et al., 2017), but all these experimental with these differences are valid and show the variation for further data.

The results which are obtained in this type of studies may show little differences due to the variation in human lumbar spine in terms of different gender, age, degeneration, experimental setup, experimental specimens (Dreischarf et al., 2014), and different type of CT scan (CT scan machine and distance between CT scan slices) as the base of studies. So the differences in their data seem to be acceptable. Moreover, due to the lack of the vivo data for IDP of the L3/L4 motion segments, we have to compare data with L4/L5 motion segments and these vivo data have a little difference with the range of its vitro data especially in the flexion and extension (Diagram 10-1).

For the post-processing step of the FE method, these data were gathered for comparing to validate the FE model data.

IDP data: according to the IDP diagram (Diagram 10-1), the flexion and extension IDP data are more similar to the minimum data of group one and two vitro data ($1\pm0.05\text{MPa}$ and $0.2\pm0.05\text{MPa}$ respectively), but they have 0.6 and 0.4 MPa differences with the flexion and extension IDP vivo data for human lumbar spine L4/L5 (in the lack of vivo IDP data for L3/L4) obtained by Wilke et al. 2001 respectively (Diagram 10-1a, 10-1b). Moreover, the lateral bending and axial rotation IDP data are more similar to the median of both the vitro (Group one and two) and vivo (Wilke et al. 2001) IDP experimental data (0.6MPa, 0.6 MPa respectively) (Diagram 10-1c, 10-1d). So all the IDP data are validated due to be in the range of vitro data.

ROM data: according to the ROM in diagram 10-2 for FE of human lumbar spine, all ROM data are lower than the vitro ROM data of group one and two in all four directions of flexion, extension, lateral bending, and axial rotation (-3 degree, -2 degree, -3 degree, and -0.5 degree respectively), but all these ROM data except flexion (-12 degree) are in the range of vivo data by Percy et al. 1984, Percy & Tibrewal 1984, and Percy 1985 (Diagram 10-2b, 10-2c, 10-2d). The same as ROM vitro data for flexion, this ROM data for flexion has more differences with the vivo ROM data by Percy et al. 1984, Percy & Tibrewal 1984, and Percy 1985 in flexion data too (Diagram 10-2a), on the other hand, all the validated vitro data are in the range of validated vivo data except in flexion which has a large differences (-8 degree). So all ROM data are in the range of validating data except ROM data in the flexion direction which the validate data has differences between its vivo and vitro data too.

Generally, the most adaptation is for axial rotation data for both ROM and IDP parts (Diagrams 10-1d, 10-2d) and the least one belongs to the flexion data in both ROM and IDP parts (Diagrams 10-1a, 10-2a). But according to the differences in gender, age, modeling software, the differences in the CT scan data, and the material properties, these little differences are normal. So all the data are validated for use in further researches.

Due to the data of IDP and ROM, these data of IDP (Diagram 10-1) are more in the range or near the mean data than ROM (Diagram 10-2). It shows that why the researcher said that the external response (ROM) is more sensitive than the internal response (IDP) to the material properties, geometry properties, and modeling software (Xu et al., 2017).

After validation, all stress and strain data (Diagram 9-1c, 9-1d) can be used to compare with different data which are obtained from prolonged analysis or testing. The maximum stress

and strain happened in inner annual fibers and the minimum happened in outer annual fibers in all four directions of flexion, extension, lateral bending, and axial rotation.

10.3 The 10 times and 100 times scaled greater of the human lumbar spine data

For analyzing the sensitivity of the FE modeling of the human lumbar spine, That FE model was scaled in its geometry and its loads by keeping the equivalent stress equal for them. So the two motion segments geometries were scaled 10 or 100 times by keeping the equivalent stress, meshing, geometry shapes, and materials equal for them, and the compression loads get 10^2 and 10^4 times and the moments get 10^3 and 10^6 times greater respectively. The result compared with each other as follow.

10.3.1 Comparing the scaled data of FE of human lumbar spine by normal size

This part contains the whole analyzed data of IDP, ROM, stress curve, and strain curve for all four positions of flexion, extension, lateral bending, and axial rotation in 10 and 100 times scaled (Diagram 10-3) according to the data which are gained from the ANSYS software (Diagrams 9-2, 9-3).

Diag 10-3: Comparing of IDP data for the structure with normal size, 10 times, and 100 times greater (a) in flexion, (b) in extension, (c) in lateral bending, (d) in axial rotation.

IDP data: these data are too similar to each other especially in 10 times and 100 times scales in all four directions, but there are little differences with a normal size which are negligible.

Diag 10-4: Comparing of ROM data for the structure with normal size, 10 times, and 100 times greater (a) in flexion, (b) in extension, (c) in lateral bending, (d) in axial rotation.

ROM data: these data are too similar to each other, especially in 10 times and 100 times scales in all four directions, but there are little differences with normal size especially in the rotation directions which are negligible.

Diag 10-5: Comparing of stress curves for the structure with normal size, 10 times, and 100 times greater (a) in flexion, (b) in extension, (c) in lateral bending, (d) in axial rotation.

Stress data: these data are too similar to each other especially 10 times and 100 times scales in all four directions, but there are little differences with a normal size which are negligible.

Diag 10-6: Comparing of strain curves for the structure with normal size, 10 times, and 100 times greater (a) in flexion, (b) in extension, (c) in lateral bending, (d) in axial rotation.

Strain data: these data are too similar to each other especially in 10 times and 100 times scales in all four directions, but there are little differences with a normal size which are negligible.

10.3.2 Validating the scaled data of FE of the human lumbar spine

For the second part of the research, by keeping the equivalent stress, material properties, and meshing parts which two motion segments have the same as normal one, the geometry was scaled 10, 100 times greater in geometry size. According to the stress definition, by keeping the stress constant, if the area is 10, 100 times greater, the compression loads have to be 10^2 , 10^4 times greater and the moment has to be 10^3 , 10^6 times greater respectively. By comparing these scaled model data with the normal size validated data which is explained before, all data for 10 and 100 times scaled are equal in each part of ROM, IDP, stress curve, and strain curve with each other, but there is a little difference (+10%, +5%, +15%, and +50% ROM, +5% in all directions of IDP, +10%, -20%, +15%, +10% in stress, and +7%, 0%, +10%, and +/-7% for strain in flexion, extension, bending, and rotation respectively) between these scaled sizes and the normal size (Diagram 10-3, 10-4, 10-5, and 10-6). These differences may occur due to the thin thickness of some parts of a normal size model such as endplates, the annular fibers, and facet joints (Xu et al., 2017) which are lose their thin thicknesses by scaling. So they have little different roles when they are scaled, but these differences are so small and negligible and they are validated due to little differences.

By scaling the geometry and the loads, according to keeping their equivalent stress equal, we received that this geometry and structure doesn't sensitive to scale and can be useful for bigger size too.

10.4 The material simplifications of human lumbar spine L3/ L4

For making the simplified structure for future modeling, we have to simplify the two motion segments in four continuous steps in order to understand the influence and effect of each material in the behavior of the two motion segments and receive the structure with fewer materials by keeping the human lumbar spine anatomical and mechanical behavior. In each step, the changes added to the previous steps not instead and then the data of ROM, IDP, stress curve, and strain curve were derived for analysis.

- Step one: omitting the annular fibers of the annular matrix which are worked just in extension.
- Step two: representing the material of annular matrix to the nucleus part of the intervertebral disc to make the integrated intervertebral disc to the human lumbar spine.
- Step three: representing the same material of annular matrix material to endplates to make a better integrated intervertebral disc.
- Step four: omitting the trabecular parts of vertebrae to make the light structure at the last step of simplification of the material part of two motion segments structure.

10.4.1 Comparing the material simplifications data by normal size data

All the results of the simplified models were based on the model under pure compression and pure moments according to its directions. Diagrams 10-7, 10-8, 10-9, and 10-10 contain the analyzed data of IDP, ROM, stress curve and strain curve for all four positions of flexion, extension, lateral bending, and axial rotation from the ANSYS software for four continuous simplified steps to compare with the analyzed data.

Diag 10-7: Intradiscal pressure (IDP) for two motion segments in four steps of material simplifications (a) in flexion, (b) in extension, (c) in lateral bending, and (d) in axial rotation.

IDP data: the result in each step is similar except between step one and step two which is the changing material of the nucleus part of the intervertebral disc.

ROM data: all the data in the four steps of simplifications are closer to each other except slight differences in the extension.

Diag 10-8: Range of motion (ROM) for two motion segments in four steps of material simplifications (a) in flexion, (b) in extension, (c) in lateral bending, and (d) in axial rotation.

Diag 10-9: stress curves for two motion segments in four steps of material simplifications (a) in flexion, (b) in extension, (c) in lateral bending, and (d) in axial rotation.

Stress data: we faced a large decrease after step one by omitting the annulus fibers in the equivalent stress of the whole two motion segments.

Diag 10-10: Strain curves for two motion segments in four steps of material simplifications (a) in flexion, (b) in extension, (c) in lateral bending, and (d) in axial rotation.

Strain curve: that the result is more similar to each other and there is a small degree difference between these steps.

10.4.2 Validating the material simplifications data

Simplifications of the materials were observed on the validated FE model of human lumbar spine L3/L4 in four continuous steps (Diagrams 10-7, 10-8, 10-9, and 10-10). According to the IDP diagrams, the result in each step is similar (less than 5% compared with the previous step) except between step one and step two (changing the material of nucleus) which have 0.4, 0.1, 0.25, and 0.24 MPa differences in directions of flexion, extension, lateral bending, and axial rotation respectively. These differences have an increase of approximately 50%, which is mainly due to changing the nucleus material and IDP declares the amount of pressure in the nucleus part (Diagram 10-7). Despite the degree of difference with FE model of this study, all these data of IDP are in the range of the experimental vivo data by Wilke et al. 2001 and vitro (group one and two) data according to the diagram 9-1. According to the ROM diagrams, all the data in the four steps of simplifications are closer to each other (± 0.05 degree near the 5 %) (Diagram 10-8). Based the similarities in ROM and IDP data in each direction except the IDP between step one and step two (which are in the range of experimental vivo and vitro data), these data are validated and shows a promising approach towards material simplifications for further structure based on the human lumbar spine L3/L4.

Based on the stress curve (Diagram 10-9), in all continuous steps except step one there are the same stress data ($\pm 5\%$) but we faced a large decrease in step one to step two by omitting the annular fibers (reduction of 500, 30, 250, and 250 MPa around -90%, -50%, -75%, and -90% for flexion, extension, bending and rotation respectively). There is the maximum stress change from the annular fiber to the cortical part of L3 and this reduction due to omitting the annular fibers. This stress reduction is good news and results in simulating method for further structure. The strain curve (Diagram 9-10) showed that the results have a variance of ± 0.4 , ± 0.1 , ± 0.1 , and ± 0.3 m/m ($\pm 40\%$, $\pm 25\%$, $\pm 35\%$, and $\pm 25\%$) with the normal FE model data for flexion, extension, bending and rotation respectively. There is the maximum strain change from the annular fiber to the annular part in all four continuous steps except the extension and lateral bending directions of solo step one which their maximum strain is in the nucleus part.

By simplification the structure of the human lumbar spine, there is the capacity of making the structure with the anatomical and mechanical role of human lumbar spine with only four main parts of cortical bone instead of whole vertebrae, annular matrix instead of a whole intervertebral disc, ligaments, and facet joints; moreover, there is a large decrease in the amount of stress. So by this method making prosthesis or making the structure based on this structure for both aspects of implanting or building joints will also reduce the time and material considerations of FE modeling.

10.5 The structural simplifications of human lumbar spine L3/ L4

After simplification based on material parts of human lumbar spine L3/L4, it is the time of simplifications based on the structure for making the elemental structure with less material and simple structure for future modeling. So we have to simplify the structure of two motion segments in four continuous steps in order to understand the influence and effect of each structure shape in the behavior of the two motion segments by keeping their anatomical and

mechanical behavior. In each step, the changes added to the previous steps not instead and then the data of ROM, IDP, stress curve, and strain curve were derived for analysis.

- Step five: simplification the main part of vertebrae to decrease the rippling surface and a more simple structure.
- Step six: simplification the posterior part of the vertebrae by keeping the scale of main part the same as the normal vertebrae and changing all the ligaments to the ALL ligaments in each side except the CL ligaments around the facet joints.
- Step seven: changing the main part of the vertebrae to the cylinder shape and the posterior part became a little wider than before to get the modular basic structure.
- Step eight: the new and final shape with cylinder vertebrae and not modular shape.

10.5.1 Comparing the structural simplifications data by normal size data

All the results of the simplified models were based on the model under pure compression and pure moments according to its directions. Diagrams 10-11, 10-12, 10-13, and 10-14 contain the analyzed data of IDP, ROM, stress curve, and strain curve for all four directions of flexion, extension, lateral bending, and axial rotation from the ANSYS software for eight continuous simplified material and structural steps.

Diag 10-11: Intradiscal pressure (IDP) for two motion segments in eight steps of material and structural simplifications (a) in flexion, (b) in extension, (c) in lateral bending, and (d) in axial rotation.

IDP data: the result in each step is more similar to the previous steps, especially with the first and second steps. But there is an increase just in step six of the extension step which has a decrease in the next steps.

(c)

(d)

Diag 10-12: Range of motion (ROM) for two motion segments in eight steps of material and structural simplifications (a) in flexion, (b) in extension, (c) in lateral bending, and (d) in axial rotation.

ROM data: all the data are more similar to each other but there is an increase in step six in all directions except the lateral bending that is changed to regular data in the next steps.

(a)

(b)

Diag 10-13: stress curves for two motion segments in eight steps of material and structural simplifications (a) in flexion, (b) in extension, (c) in lateral bending, and (d) in axial rotation.

Stress data: we faced a large decrease in step one by omitting the annulus fibers in the equivalent stress of whole two motion segments in all directions except extension, and the most stress after the normal size data belongs to the step five by changing the main part of the vertebrae.

Flexion stress curve: all the data are between 5.15×10^8 Pa (the normal structure with annulus fibers) and 1.8×10^8 Pa (the new structure which is not modular) the maximum is at first on annulus fiber then converted to the L3 vertebrae after omitting the annulus fibers.

Extension stress curve: all the data are between 1.6×10^8 Pa (the normal structure by changing the main part of vertebrae) and 0.4×10^8 Pa (the new structure which is not modular) the maximum is at first on annulus fiber then converted to the L3 vertebrae after omitting the annulus fibers.

Lateral bending stress curve: all the data are between 3.1×10^8 Pa (the normal structure with annulus fibers) and 1.2×10^8 Pa (the new structure which is not modular) the maximum is at first on annulus fiber then converted to the L3 vertebrae after omitting the annulus fibers.

Axial rotation stress curve: all the data are between 2.75×10^8 Pa (the normal structure with annulus fibers) and 0.3×10^8 Pa (the normal structure which have the same material for endplate, annulus, and nucleus) the maximum is at first on annulus fiber then converted to the L3 vertebrae after omitting the annulus fibers except for the changing the main part of vertebrae steps, the maximum stress was on the L4 vertebrae.

Diag 10-14: Strain curves for two motion segments in four steps of material simplifications (a) in flexion, (b) in extension, (c) in lateral bending, and (d) in axial rotation.

Strain curve: The first four steps have more strain than the second four steps in all directions except the extension part, all the strain data were decreased, and there is a small degree difference between these steps.

Flexion strain curve: all the data are between 1.8 mm (the normal structure with omitting trabecular) and 0.2 mm (the normal structure with changing the main part of vertebrae) the maximum is at first on annuals fiber then converted to the annuals after omitting the annuals fibers then converted to the intervertebral disc after having the same material for annuals, endplates, and nucleus.

Extension strain curve: all the data are between 0.25 mm (the normal structure with omitting trabecular) and 0.1 mm (the new structure with modular shape) the maximum is at first on annuals fiber then converted to the nucleus after omitting the annuals fibers then converted to the intervertebral disc after having the same material for annuals, and nucleus then converted to the facet after changing the main part of vertebrae.

Lateral bending strain curve: all the data are between 0.9 mm (the normal structure with annuals fibers) and 0.2 mm (the new structure with modular shape) the maximum is at first on annuals fiber then converted to the nucleus after omitting the annuals fibers then converted to the intervertebral disc after having the same material for annuals, and nucleus except for changing the main part of vertebrae which are the maximum was on facet.

Axial rotation strain curve: all the data are between 0.919 mm (the normal structure with omitting trabecular) and 0.127 mm (the normal structure with changing the main part of vertebrae) the maximum is at first on annuals fiber then converted to the annuals after omitting the annuals fibers then converted to the facet after changing the main part of vertebrae.

10.5.2 Validating the structural simplifications data

Simplifications of the structure were observed on the validated FE model of human lumbar spine L3/L4 in four continuous steps more than the previous four steps of material simplifications (Diagrams 10-11, 10-12, 10-13, and 10-14). According to the IDP diagrams, the result in each step is similar to each other (+20%, +30%, +5%, and +5% for flexion, extension, bending, and rotation respectively) except in the extension part in step six by changing the posterior part of the vertebrae which has 0.278 MPa differences in extension ($\pm 50\%$) with the previous step. This difference was happened because of the thin thickness of joints between the posterior and anterior part in a new simplified structure and these differences in next steps was corrected by changing its thickness. Despite the degree of difference with FE model of this study, all these data of IDP are in the range of the experimental vivo data by Wilke et al. 2001 and vitro (group one and two) data according to the diagram 10-1, so by having these differences, the IDP data are validated. According to the ROM diagrams, all the data in the four steps of simplifications are closer to each other ($\pm 30\%$, +300%, -30%, and -20% for flexion, extension, bending, and rotation respectively) except slight differences between step five and six by changing the posterior part of vertebrae which were corrected in next step by changing the thickness of the joint between posterior and anterior part. These differences are 0.51, 0.394, -0.04, and 0.3 degrees for flexion, extension, lateral bending, and axial rotation respectively ($\pm 50\%$, $\pm 600\%$, -5%, and $\pm 45\%$ for flexion, extension, bending, and rotation respectively). These differences are a little and they are in the range of both the experimental vivo data by Percy et al. 1984, Percy & Tibrewal 1984,

and Pearcy 1985 and vitro (group one and two) data of human lumbar spine so they are negligible (Diagram 10-8). Based the similarities in ROM and IDP data with experimental data in each direction except the IDP between step five and step six which are minute and in the range of experimental vivo and vitro data too, these data are validated and shows a promising approach towards material simplifications. Based on the stress curve (Diagram 10-13), we faced a decrease from step one for all steps except step five which are changing the main part of vertebrae. The stress curve shows the decrease in stress amount in all four directions (-80%, -65%, -65%, and -80% or flexion, extension, bending, and rotation respectively). There is a change in the maximum stress from the annular fiber to the cortical part of L3 in all eight continuous steps except the axial rotation of step five which the maximum stress happened in the L4 cortical part. The strain curve (Diagram 10-14) shows that the result is more similar to each other and there is a small degree difference between these steps and the four-step structural simplification have the least strain data and it face decrease compared with the material simplification data (-80%, -50%, -65%, and -75% or flexion, extension, bending, and rotation respectively). The maximum strain in four steps of structural simplifications are in the facet in the extension, lateral bending, and axial rotation, and in the intervertebral disc in the flexion of all-new four steps and step six lateral bending data directions.

By simplification the structure of the human lumbar spine, there is the capacity of making the structure with the anatomical and mechanical role of human lumbar spine but only with four main parts of cortical bone with new structure instead of whole vertebrae, intervertebral disc with the annular matrix character, two type of ALL and CL ligaments, and facet joints.

10.6 The 10 times and 100 times scaled greater of new structure

For making the structure based on the new structure which is made based on two motion segments of the human lumbar spine, applying features of the new structure in normal size with changing the scale of that structure and it's loads by keeping the equivalent stress equal for them. So the new structure geometries were scaled 10 or 100 times by keeping the equivalent stress, meshing, geometry shapes, and materials equal for them, and the compression loads get 10^2 and 10^4 times and the moments get 10^3 and 10^6 times greater respectively.

10.6.1 Comparing the scaled data of new structure by normal size data

This part contains the whole analyzed data of IDP, ROM, stress curve, and strain curve for all four positions of flexion, extension, lateral bending, and axial rotation in 10 and 100 times scaled of the new structure based on human lumbar spine (Diagram 9-12, 9-13) according to the data which are gained from the ANSYS software (Diagrams 10-15, 10-16, 10-17, and 10-18).

Diag 10-15: Comparing of IDP data for the new structure with normal size, 10 times, and 100 times greater (a) in flexion, (b) in extension, (c) in lateral bending, (d) in axial rotation.

IDP data: these data are too similar to each other especially in 10 times and 100 times scales in all four directions, but there are little differences with a normal size which are negligible.

Diag 10-16: Comparing of ROM data for the new structure with normal size, 10 times, and 100 times greater (a) in flexion, (b) in extension, (c) in lateral bending, (d) in axial rotation.

ROM data: these data are too similar to each other especially in 10 times and 100 times scales in all four directions, but there are little differences with normal size too which are negligible.

Diag 10-17: Comparing of stress curves for the new structure with normal size, 10 times, and 100 times greater (a) in flexion, (b) in extension, (c) in lateral bending, (d) in axial rotation.

Stress data: these data are too similar to each other in 10 times and 100 times scales in all four directions, but there is a large decrease in all four directions with the normal size.

Diag 10-18: Comparing of strain curves for the new structure with normal size, 10 times, and 100 times greater (a) in flexion, (b) in extension, (c) in lateral bending, (d) in axial rotation.

Strain data: these data are too similar to each other especially in 10 times and 100 times scales in all four directions, but there are little differences with a normal size, which are negligible.

10.6.2 Validating the scaled data of new structure

By keeping the equivalent stress, material properties, and meshing parts for new structure based on two motion segments of the human lumbar spine the same as normal one, the geometry was scaled 10, 100 times greater in geometry size the same as what happened in chapter 7 scaling. According to the stress definition, by keeping the stress constant, if the area is 10, 100 times greater, the compression loads have to be 10^2 , 10^4 times greater and the moment has to be 10^3 , 10^6 times greater respectively. By comparing these scaled model data with the normal size validated data which is explained before, all data for 10 and 100 times scaled are equal in each part of ROM, IDP, stress curve, and strain curve with each other, but

there is a little difference between these scaled sizes and the normal size (Diagram 10-15, 10-16, 10-17, and 10-18) (+20%, +20%, +20%, and +7% IDP, +15%, +25%, +25%, and +25% ROM, +20%, +20%, +15%, and +15% in strain for flexion, extension, bending, and rotation respectively) especially in their stress data (-90%, -75%, -25%, and -75% for flexion, extension, bending, and rotation respectively). These differences may occur due to the thin thickness of some parts of a normal size model (Xu et al., 2017) which are loosen their thin thicknesses by scaling and can distribute the loads better. So, they have little different roles when they are scaled, but these differences are so small and negligible. So these structures are validated due to little differences especially in scaled sizes and would be useful for a bigger size structure because of having the ability of tolerance of the much loads for each position, facing the decrease in stress amount, and having the other role of human lumbar spines such as static role, dynamic role, modular form, and lightweight structure. So, in this research, we showed that these new structures have the potential of making the building joints (10 times scales) and building structure (100 times scaled) by tolerance these large amounts of loads.

10.7 Architectural samples based on the new structure

All the above sections contain the validation of the FE models to receive the final FE of the new structure, which are useful for building near-future systems. So, this new structure was validated to be useable for future structures by having the ROM and IDP data in the range of experimental vitro and vivo data by having the less stress and strain amount compared with the two human lumbar spine motion segments.

According to the chapter four, making the structure based on the human spine structure cause to have a structure, which are so light, stable while moving, and modular form by having the complex roles of the human spine character and being resistance against the internal and external response such as earthquake or wind (Picture 10-1). At this part, there is a description and show of the samples, which have the capacity of making by this new structure.

By being this structure modular role, it can be repeated up to each other to get different high-rise building (Picture 10-2, 10-3).

Having the capacity of validation in scaled greater; it has the capacity of scaling in different sizes based on the amount of the space. Moreover, these scaled sizes can be repeated up to each other too (Picture 10-4).

The most important role of this structure is having the capacity of making structure in a different form of human body roles such as crawling, leaning, jumping, and etc. by repeated that new structure in a different form and directions for making the unusual buildings or bridges.

NEW BUILDING STRUCTURE SAMPLE 2 Perspective

Pic 10-3: The new building structure sample 2 perspective

■ FRONT ELEVATION

■ LEFT ELEVATION

■ BACK ELEVATION

■ SECTION

■ NEW BUILDING STRUCTURE SAMPLE 3

CHAPTER XI: CONCLUSIONS

11. Conclusions

The main conclusion of modeling and the other general conclusions of the study based on the results, which are carried out experiments, and tests are presented in this chapter.

11.1 The main conclusions of modeling

Throughout this study making the elemental structure for building joints and building structure based on the anatomical and mechanical behavior of the human lumbar spine (L3/L4) have been investigated by the FEM. This FE of the human lumbar spine (L3/L4) was made based on CT scan data as the base of the vertebrae geometry, verified geometry data for making the other geometry manually based on CT of vertebrae, verified material data, and the combination loads of pure compression and pure moments in four directions of flexion, extension, lateral bending, and axial rotation. After making the FE of the human lumbar spine, the model was checked for Intradiscal pressure and Range of motion data by comparing with validate experimental vivo and vitro data and scaling it 10 and 100 times greater for testing its sensitivity and validating the FE model. Then, the human lumbar spine was simplified in material parts in four continues steps of 1- omitting the annuals fibers, 2- representing the annuals matrix material to the nucleus part, 3- representing the annuals matrix material to the endplates, and 4- omitting the trabecular part of the vertebrae to make the two motion segments simple for further modeling based on that with fewer materials. Moreover, the FE model was simplified again in four continues steps (in its structure) of 1- simplifying the main part of the vertebrae, 2- simplifying the posterior part of vertebrae, 3- changing the main part of the vertebrae to a cylinder shape, and finally 4- simplifying the new structure to not modular shape. Their data were checked for a range of motions (ROM) and intradiscal pressure (IDP) by the validated experimental vivo and vitro data to analyze their internal and external response and their stress curves and strain curves for both material and structural simplification steps were obtained to check these steps of simplifications. At the end, the FE model was scaled 10 and 100 times greater again by keeping its equivalent stress, mesh, material, and the form of geometry equal to the new structure, so the pure compression becomes 10^2 , 10^4 and the pure moment becomes 10^3 and 10^6 greater for these 10 times and 100 times scaled greater respectively due to keeping the equivalent stress equal with the normal size of new structure to analyze and check its sensitivity for bigger size structures. Thus, according to the analyzed data which are documented in this thesis and are in the range of the both IDP and ROM experimental vivo and vitro data, the modeling method introduced in this study was proved to be valid and was demonstrate the accurate building of simple structure based on the anatomic and biomechanics of the human lumbar spine with just five material of 1- cortical bone instead of the whole vertebrae with empty place inside, 2- the annuals matrix material instead of the whole intervertebral disc, 3- ALL and CL ligaments, and 4- facet joint material; moreover, there is a decrease in the number of stress and strain in the whole two motion segments of the new structure compared with the two motion segments of human lumbar spine.

11.2The General conclusions

Considering the objectives of this study, the general conclusions arising from this work, which are described more in their relevant chapters, are summarized as follows:

11.2.1 General conclusions of bionics part

In response to the minor objectives, these conclusions have been received:

- Bionic science deals with reviewing, studying of the structures and patterns in nature, applying them to solve human problems, and finding a solution to the ecological crisis based on the knowledge derived from nature has two approaches of problem-oriented approach and solution-focused approach. Due to bionic has long evolution journey over millions of years, the more similar our performance is to nature, the more guarantee for our longer survival.
- Bionic architecture as the main part of bionic science tries to design and construct the buildings based on layout and lines of natural forms in three interactions types of architecture of nature, architecture with nature, and architecture from nature. Bionic architecture matures in the early 21st century by dealing with the combination of biology, mathematics, construction, and mechanics.
- Bionic architecture which is in the works of Leonardo Da Vinci, Greg Lynn, Charlie Luxton, Paul Klee, Tadao Ando, Frank Lloyd Wright, Le Corbusier, R. Fisher, Santiago Calatrava Valls, Norman Robert Foster, etc. mimics in three levels of the organism level, the behavior level, and the ecosystem level; moreover, it classifies to six types of Stencils, Structure, Materials, Form, Function, and Process. The previous architectures structures were made more by the natural Stencil and Form, less by the natural Function and Structure, and the natural Material and Process were negligible.
- Nature is an unlimited source of inspiration and a good teacher in various fields of knowledge, technology, science, and architectural design for the bionic architecture field. The forms of human relations and interaction with nature during the human life are categorized to Nature and organic model (the Hunting Age or Stone Age), Nature and hyper organic model (the farm age), Nature and dominate nature model (the industrial revolution), and Nature and industrial quasi-organic model (Nowadays).
- The human body as a part of nature is a good example for the near-future buildings because of its responsibility for efficiently and dynamically behavior that has to force the load placed upon them and its ability to be re-shaped. The human body the same as the other natural systems is consist of a number of subsystems such as the human skeletal joints (to make the frame of the human body), the human brain and nervous system (to coordinate actions between differentiated cells and organs of the body), and the human muscles (to provide stability and ability to do many functions such as moving, breathing, running, walking, writing, dancing, etc.).

- The human body by having the best shape, the best features and characters, aesthetic form, comfort form, dynamic form, changeable form, and light structure is a good solution for building problems.
- Each human joints have different flexibility and actions such as flexion, extension, abduction, adduction, rotation, and circumduction, which are useful for, further modeling actions too.

11.2.2 General conclusions of the skeletal system part

In response to the minor objectives, these conclusions have been received too:

- The human joints are categorized and described based on their movement, structure, the degree of freedom; shape, biomechanical classification, and anatomical classification, which are useful for further modelings, depend on their aims. Two categories of human joints are so useful for modeling such as:
 - Joint types by their function and motion: immovable joints, slight movement joints, and full movement joints.
 - Joint types by their structure: fibrous joints by three subsets, cartilaginous joints by two subsets, and synovial Joints by six subsets.
- The human joints based on their structure and functions are used in different field of industrial design, building structures, medical prosthesis or implanting consciously or unconsciously which are described a lot in chapter three. For example the fibrous joints (no movement joint) were used to design the fixed joints in building or industrial, the cartilaginous joints (some movement joint) were used to make the joints and building structure with a little movement, and the synovial Joints (full movement joints) were used to make the moveable, portable, ratable joints, buildings, and bridges.
- The human skeletal system is the framework of the body and is gathered from bones (framework of body), cartilage (a smooth well-lubricated surface), synovial fluid (to slide motion freely), discs (a shock absorber), and ligaments (to join the bones together) which are compacted to each other in human joints.
- The human spine as the main joint of the human body makes our body structure, body-supporting, body-standing, body-up righting, body movement, body twisting freely, body-bending with flexibility, and protecting our spinal cord. The spinal column has four main sections-the cervical spines (The first seven vertebrae), the thoracic spine (The mid-back consist of 12 vertebrae), the lumbar spine (usually five vertebrae), and sacrum (five vertebrae) with coccyx (four vertebrae) which have special characters in the human body functions. So having this role of stability and mobility at the same time without collapsing, doing more roles, and bearing much loads cause to make building structure based on that structure.
- The human spine has contained the vertebrae (cortical, trabecular), intervertebral disc (annual matrix, annuals fibers, nucleus, and endplates), ligaments (ALL, PLL, CL, TL, ISL, SSL, and

LF), and facet joints as an anatomical aspect. These parts have special characters and functions in the anatomical and mechanical behavior of the human spine.

- The human lumbar spine where the most weight bearing is happened by having beneficial features of aesthetic form, the complex roles of different functions, stable in different parts, the ability to move and change to be balanced, the modular shape, and lightweight structure is a good structure for future modeling.

11.2.3 General conclusions of the anatomical roles of the human spine part

In response to the major objectives, this conclusion has been received:

The human lumbar spine can convert to the simple structure by its anatomical behavior, which is tolerating against the internal and external loads. This way of converting was explained entirely in chapter four. The products of this chapter was the building structure by the anatomical roles of the human lumbar spine with vertebrae, intervertebral disc, ligaments, and facet joints. There is a capacity of making the simple structure moveable by adding muscles roles and making conscious by adding nerves systems roles to the designed structure.

11.2.4 General conclusions of the mechanical roles of the human spine part

In response to the minor objectives, these conclusions have been received too:

- Each part of the human lumbar spine has special mechanical behavior and properties such as its material compositions, its characters, its scale and dimension, which are affected by the age, gender, disease, drying, embalming, and race. Setting the main chart of this mechanical behavior such as element type, element number, material formulation, material Modulus of elasticity and Poisson's ratio from the verified data happened during the chapter five for better modeling by software or making prototype for further analyzing of these structures (Table 6-1 and 6-2).

- For modeling based on the mechanical part of the human lumbar spine, we have to change a little mechanical behavior of some parts according to validate data to get the better model by software. Usually, spine materials are non-homogeneity, anisotropy, and nonlinearities, but for making analyze easily they are assumed linear, isotropic and with elastic orthotropic structure, and homogeneous.

- Modeling based on human lumbar spine needs the number of loads in different directions. The human lumbar spine is under pure compression due to the local muscle forces and upper body weight and pure moments due to the directional movements of flexion, extension, lateral bending, and axial rotation. These both pure compressions and pure moments' amount depend on their directions too. So, the mixed of compressive force and moment of 1175N and 7.5Nm, 500N and 7.5Nm, 700N and 7.8Nm, and 720N and 5.5Nm are applied for flexion, extension, lateral bending, and axial rotation, respectively (Table 6-3).

- Validating the model is acquired by comparing its data with validated IDP (the pressure in the nucleus part) and ROM (the degree differences between two vertebrae after and before

loadings) data. So, for validating the model of the human lumbar spine model, the experimental vivo and vitro validate data for IDP and ROM were gathered and were set as charts for comparing (Diagrams 10-1 and 10-2). Being in the range of the vitro or vivo experimental data for IDP and ROM shows the validation of the research data.

11.2.5 General conclusions of the finite element approach part

In response to the minor objectives, these conclusions have been received too:

- Making model happened in two ways of experimentation based on human, animal, natural form, artificial, cadaveric, or prototype models and finite element method. Finite Element Method (FEM) is a numerical and a computer-based analysis tool by using mathematics to simulate and analyze engineering products and systems in a virtual environment, which is used by engineers and scientists in a wide range of industries such as automotive, biomechanical industries, and aeronautical. The finite element method involves modeling the structure by using small interconnected Elements which are connected to one another in Nodes to make a complete set of Meshes.
- Developing a model by the FEM happens in three main steps of pre-processing, processing, and post processing steps. Moreover, each step may contain sub-step according to the goal of studying and modeling. Each step needs a special software or compound of software for modeling too.
- Modeling the human lumbar spine by software needs some special mechanical behaviors, which are different from normal one to model with a better effect. These mechanical behaviors for the geometry or material of each part of the human lumbar spine for modeling by the software were described a lot during chapter six and seven.
- Modeling the human lumbar spine by the FE method has these benefits as follow:
 - Prediction of the biomechanics of the human lumbar spine by software.
 - Determining the role of material properties and the behavior of the spine.
 - Performing a parametric analysis of loadings (stresses in the discs, vertebrae, and ligaments and very detailed motion data) and comparing it to experimental studies.
 - Studying the load transfer mechanisms and load distributions among different spinal components.
 - Assessing the spinal performance of disease, degenerative changes, trauma, aging, and surgery, damaged or healthy spine.
 - Assisting in the design and development of new spinal implants.
 - Handling the various geometries and materials of the prosthesis and examined them if required.
 - Changing the geometries to test it easily.

- Doing different types of analysis repeatedly without damaging or changing in our models.
- Measuring the internal loads of the spine.
- Getting the internal and external response of each geometry under different loadings.
- Making the basic model for the implants, prosthesis, and structures, treating the lumbar diseases, and modeling of the human lumbar spine.

11.2.6 General conclusions of modeling the human lumbar spine by FEM part

In response to the major objectives, these conclusions have been received too:

- Modeling the two motion segments of the human lumbar spine was happened based on CT scan data of human lumbar spine of a normal person as a base of geometry and verified data. This process needed different and compound of software such as 3d-slicer, MIMICS, 3D-matic, Rhino, ANSYS (material book, space claim part, mechanical part) in steps of making geometries, representing materials, loading the bars in different directions, choosing solutions, and getting its data for validating and checking its sensitivity. All the IDP data are validated due to be in the range of vitro data. All ROM data are in the range of validating vivo data except ROM data in the flexion direction (-12 degree) which the validate data has differences between its vivo and vitro data too (-8 degree). So The FE of the human lumbar spine by comparing with the experimental vivo and vitro data was validated.
- Checking the FE of human lumbar spine sensitivity for the bigger size for testing its capacity for making building joint and building structure. It was scaled 10 and 100 times greater by keeping its stress and mesh amount equal with the normal size, but the compression loads were scaled 10^2 and 10^3 greater and the moments were scaled 10^4 and 10^6 greater. All data for 10 and 100 times scaled are equal in each part of ROM, IDP, stress curve, and strain curve with each other, but there is a little difference (+10%, +5%, +15%, and +50% ROM, +5% in all directions of IDP, +10%, -20%, +15%, +10% in stress, and +7%, 0%, +10%, and +7% for strain in flexion, extension, bending, and rotation respectively) between these scaled sizes and the normal size. These differences may occur due to the thin thickness of some parts of a normal size model such as endplates, the annular fibers, and facet joints, which are, loosen their thin thicknesses by scaling. So, these structures have the capacity of making greater scale by tolerating much loads.
- By scaling the geometry and the loads, according to keeping their equivalent stress equal, we received that this geometry and structure doesn't sensitive to scale (the same data for both 10 and 100 times greater) and can be useful for bigger size too.

11.2.7 General conclusions of simplifying the FE of the human lumbar spine part

In response to the major objectives, these conclusions have been received too:

- Designing the simple structure for implanting, making prosthesis, or knowing the mechanical behavior of each material in internal and external response was happened by simplification the material part of the human lumbar spine in four continues steps. According to the IDP diagrams, the result in each step is similar (less than 5%) except between step one and step two (an increase of approximately 50%) which is mainly due to changing the nucleus material and IDP declares the amount of pressure in the nucleus part. Despite this difference, all IDP data are in the range of the experimental vivo and vitro data. According to the ROM diagrams, all the data in the four steps of simplifications are closer to each other (± 0.05 degree near the 5 %). So these data are validated and shows a promising approach towards material simplifications for further structure based on the human lumbar spine L3/L4 just with the cortical bone, annulus of the disc, 7 types of ligaments, and facet joint materials. Moreover, the stress curve showed a large decrease between step one to step two (reduction of around -90%, -50%, -75%, and -90% for flexion, extension, bending and rotation respectively) by omitting the annulus fibers which is good news and result in simulating for further structure. Moreover, the strain curve results have a variance of $\pm 40\%$, $\pm 25\%$, $\pm 35\%$, and $\pm 25\%$ with the normal FE model data for flexion, extension, bending and rotation respectively

- Creating the elemental and basic structure based on the human lumbar spine was happened by structural simplification more than the simplification of the material for making further joints in industrial design or building structures. According to the IDP diagrams, the result in each step is similar to each other (+20%, +30%, +5%, and +5% for flexion, extension, bending, and rotation respectively) except in the extension part in step six by changing the posterior part of the vertebrae ($\pm 50\%$) due to the thin thickness of joints between the posterior and anterior part in a new simplified structure which was corrected by changing its thickness in next step. Despite this difference, all these data of IDP are in the range of the experimental vivo data and vitro data and are validated. According to the ROM diagrams, all the data in the four steps of simplifications are closer to each other ($\pm 30\%$, +300%, -30%, and -20% for flexion, extension, bending, and rotation respectively) except slight differences between step five and six ($\pm 50\%$, $\pm 600\%$, -5%, and $\pm 45\%$ for flexion, extension, bending, and rotation respectively) by changing the posterior part of vertebrae which were corrected in next step by changing the thickness of the joint between posterior and anterior part. These differences are a little and they are in the range of both the experimental vivo and vitro data so they are negligible. The stress curve shows the decrease in stress amount in all four directions again (-80%, -65%, -65%, and -80% or flexion, extension, bending, and rotation respectively) and the strain curve faces decrease compared with the material simplification data (-80%, -50%, -65%, and -75% or flexion, extension, bending, and rotation respectively). So this simplification data are validated and shows a promising approach towards structural simplifications with a simple form and just with the cortical bone, annulus, two types of ligaments, and facet joints material with a decrease in stress amount in all four directions.

- Testing the sensitivity of the new structure was happened by scaling the geometry of the human lumbar spine and new structure based on the human lumbar spine 10 times or 100 times greater and getting their data for further analysis. All data for 10 and 100 times scaled are equal in each part of ROM, IDP, stress curve, and strain curve with each other, but there is a little difference between these scaled sizes and the normal size (+20%, +20%, +20%, and

+7% in IDP, +15%, +25%, +25%, and +25% in ROM, +20%, +20%, +15%, and +15% in strain for flexion, extension, bending, and rotation respectively) especially in their stress data (-90%, -75%, -25%, and -75% for flexion, extension, bending, and rotation respectively). These differences may occur due to the thin thickness of some parts of a normal size model, which lost their thin thicknesses by scaling and can distribute the loads better. So these structures are validated due to little differences especially in scaled sizes and would be useful for a bigger size structure because of having the ability of tolerance of the much loads for each position, facing the decrease in stress amount, and having the other role of human lumbar spines such as static role, dynamic role, modular form, and lightweight structure.

- At the end of this simplification study, by the deformation, ROM, IDP, and scaled data we showed that these new structures by simplification of the material and structural part of human lumbar spine (L3/L4) are good for modeling the future joints (10 times scales) and building structures (100 times scaled) by tolerance these large amounts of loads, being their data in the range of the experimental vivo and vitro data, having the less stress, and having less material than normal one sample.

- As a final conclusion, this architectural structure was made just based on the architectural bionics, which tries to imitate the nature to make the new structures and make sense to them. There are much more natural form, which can imitate in different levels to make new structures for future sensible structures.

CHAPTER XII: FUTURE RESEARCH LINES

12. Future research lines

Further recommendations for future studies in related fields are presented in this final chapter.

The relation between buildings and nature starts from the beginning of human life, the challenges of making buildings are solved by having relation with nature like natural shelters, and archetypes such as caves and trees to model the mechanical, electrical, structural, protection, cooling, shell building and security, but humans used nature and environment unfeeling and affected adversely the planet's natural balance by causing climate changing and global warming. Bionics structures caused a revolution in the development of technology nowadays, and it is still being developed (Taghizadeh & Bastanfard, 2012) by the combination of biology, construction, mathematics, mechanics, and computational programs and software to cause the sustainable development of complex systems by increasing their efficiency (Salsabili et al., 2018, 2016).

There are extensive examples in nature for modeling and many of them have already been used (Salsabili et al., 2018, 2016) in a different field of studying such as industrial design or bionic design, but using these examples in the architecture and building are too less and only based on their forms. Knowing their anatomical and mechanical structures of these examples of natural causes to make a sustainable and adaptable structure for further modeling. The more our designs based on the natural structures, the more they will be adaptable with nature. The FE helps to model these structures easily, cheaper, faster, and more beneficial than the other ways and can change or simplify these natural forms to get better results. So this study is the example of making a bridge between nature and buildings by using both anatomical and mechanical aspects of human joints.

- As I mentioned, there are a different type of joints in the human body which are used consciously or incautiously in joints or building, by FE method we can simulate them and using these joints for further modeling in the different field of studying too.
- By recognizing the spine joints and its curves, it is possible to simulate different forms and states of the human body such as lying, jumping, sitting, bending, etc. at different ages of infancy, childhood, youth, aging, weightlifting, etc. with different curves for future requests by having both aspects of stability and mobility in different field of studying.
- In this thesis, I explained one way of simplification of the human lumbar spine to get the structure, but there are more than this way of simplification of that structure. By the FE method, we can check different way of simplifications based on our desire for our modeling and check their influence easily.
- Each material of human joints have specific mechanical and anatomical behavior, by knowing this behavior and finding the best materials, there is the capacity of substituting them with these human spine materials to make the better model for further structures.
- There is the capacity of making a prosthesis, implants, joints, and any type of structures in different field such as buildings, industrial design by these type of joints by this type of

mechanical and anatomical behavior and having the capacity of loading these much loads by its light structure.

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