CHAPTER IV
FINITE ELEMENT ANALYSIS OF THE KNEE JOINT WITHOUT A MEDICAL IMPLANT

4.1 Modeling in Biomechanics

The human body, apart of all its other functions is a mechanical mechanism and a structure, since it transmits movement and force. As a structure its elements may be subjected to forces and suffer stress and strain. Biomechanical analysis has been made for many years, and it has been a huge effort to analyze theoretical and structural behavior of components of the human being when subjected to a force.

This work has been carried out by investigators of many areas; biologists, doctors, physicists, chemistries, and of course engineers. But before starting with the analysis is important to consider the principles of the science of analyze stress and strain. Human muscle – skeletal system consists of a great number of interconnected systems, muscles, tendons, bones, cartilages, veins, etc. Each one of them provide and are subjected to external forces, the all human body generates force. [33]

In order to solve a stress / strain analysis it is important to consider some parameters, like the known and unknown forces, stress, strain and known and unknown displacement. These parameters also interconnected by the equilibrium, which involves forces and stresses, mechanical properties, referring stress and strain and compatibility relating strain and displacement as shown in Fig. 4.1.

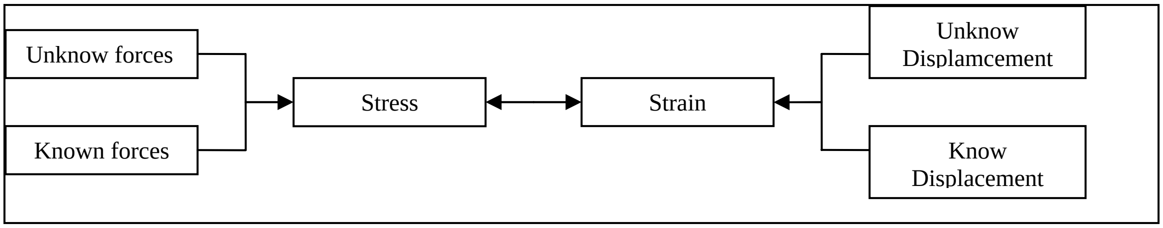

Fig 4.1 The structural chain

It is important to consider that a stress analysis in human components in not as simple as it sounds in theory. In practice there are several other parameters that have to be taken into account. For example, there are infinite number of points on the boundary where are presented displacements and there are forces acting. Somehow this can be seen as having an infinite number of equations with an infinite number of unknowns. [33]

So, to reduce the problem to a finite number of solutions, it is necessary to simplify the problem by making assumptions. These are the most common assumptions used to simplify the problem: [27]

1. Geometry: Draw a simpler shape of the human component than it really is.
2. Material: Behavior of the material under certain conditions is more normal than it really is.
3. Form of material: Inhomogeneity and anisotropy can be ignored.
4. Structure distribution: Consider it as continue, for example the human bone is porous, consider it as a solid structure.
5. Boundary conditions: Consider them simpler than they really are.
6. Form of solution: Arbitrarily assumed, for example from experience.
7. Mathematics: used to get a solution from data given.

4.1.1 Applying Biomechanics

In usual engineering problems is not quite usual obtain a precise and efficient solution. In biomechanical stress analysis the thing is even more complicated, only to obtain the basic data for the model, geometry, properties, boundary conditions, it is such a challenge. Anatomic structures are quite different from each other for different persons. Data must be obtained by medical studies and even like this is very difficult to get a precise result.

Muscles are an essential structural component of the body, they carry load, but also as a living constituent they also create tensile forces. Then it is needed to make a free body diagram of the element to analyze the internal forces coming from muscles and tendons. Therefore the calculation of higher amount of stresses in muscle-skeletal system is a special case of structural analysis because that involves techniques not required for conventional engineering. In this study we will concentrate on the knee joint. [6]

4.2. Bone properties

As a first step to realize a finite element analysis to knee joint, is important to know the properties of the joint, starting with the main part of it, bones. As it has been seen before, this joint consists of three main parts: tibia, lozenge and femur, two of them are one of the larger bones of human been.

As a mater of fact, bone tissue is neither homogeneous nor isotropic morphologically. Whether it can be regarded as homogeneous with respect to its mechanical properties or not remains to be seen.

According to Young the tensile stresses in bone are resisted by the collagen, and the compressive stresses by the apatite. Further, Currey reports that bone as a brittle substance has a very high tensile strength in relation to its compressive strength.

Knese, Currey, Young, they all did tests in human bones to determine their mechanical properties, and the three of them despite the fact that their methods where different, they have something in common, they did tests with “dry” and “wet” bones. The difference is that dry bone is an unphysiological procedure, and this alteration was studied by several authors, and has been found a considerable increase in strength after drying. But with wet bones we get a better approximation of the real behavior of bones in human being, so well take wet bones properties obtained from the investigation made by Rauber in 1876, determining that a fracture in bone is presented at 260 GPa.

Bones are part of the muscle skeletal system, bones are supposed to transmit load by the application of muscle forces. As we want to simulate a physiological distribution of strain and stress in bone, this experiment should be intended to simulate the application of natural load balanced by tension forces applied from the muscles and tendons. [45]

4.3. Loads applied

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Weight is the primary force and contact forces are the external reaction to this load. Contact forces are for example, while standing, the weight forces are balanced by the ground reaction, loads applied by the ground through the soles of the feet. This load is caused by gravity force which is applied throughout the volume of the body; each part of the body adds its own contribution.

While standing the resultant of the ground reaction forces and the sum of forces applied to both feet is the same than the weight of the body. This is supposed to be in equilibrium by the ground reaction and the body weight. The sum of forces in lower leg is supposed to be zero so that the structures of the knee combine to provide a force which exactly balances the ground reaction.

This is the resultant force F which resists translation of leg. It is equal in magnitude and opposite in direction to the ground reaction W. The resultant force passes through point O. This line of action does not coincide with the line of action of W, although the two forces are parallel, so the knee must therefore combine to provide a moment C that resists the rotary movements of lower leg. Magnitude of this moment C must be equal to the product of the force W times the perpendicular distance *d* from O to W. Fig. 4.2

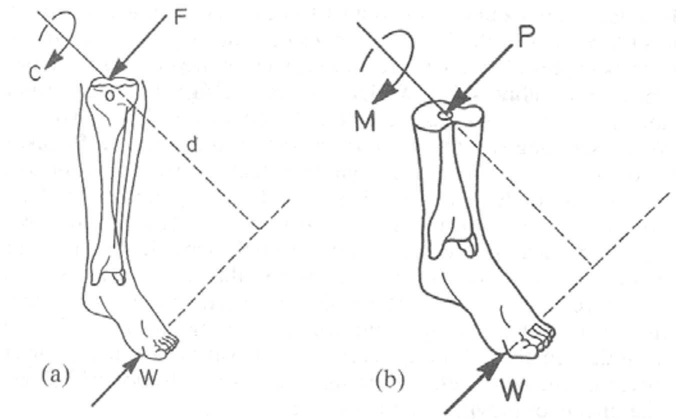

Fig 4.2. a)Lower leg in equilibrium b)Section through the tibia showing resultant force P and resultant moment M needed for equilibrium under the action of external load W.

The purpose of this study is first to analyze the stress and strain concentration during a gait cycle which is popularly known as a step, this cycle begins when foot strikes the ground and ends when the same foot strikes the ground again. The gait cycle has two main parts: The stance phase and the swing phase. These stages are shown in figure 4.3. [95]

Stance phase occurs at the beginning of the cycle and occurs during 60% of it; it has two periods, 10% each, when the center of gravity is at its lowest and it is divided into 5 phases:

1. Initial Contact: At this point the knee is extended and the ankle is neutral, or plantarflexed. Usually the heel contacts the ground first.
2. Loading response: This is the first period of double limb support and ends at contralateral toe off, when the opposite foot leaves the ground. At this

point, knee flexes by 15 degrees and quadriceps act to maintain hip and knee stability.

3. Mid stance: It begins with contralateral toe off, and ends when the knee is extended and the ankle is neutral .
4. Terminal stance: It begins when the knee is totally extended and perpendicular to the ground and it ends when the contralateral foot contacts the ground. Terminal stance and mid stance are the only phases when the center of gravity truly lies over the base of support.
5. Pre swing: This phase begins at the second double support; then the contralateral initial contact and ends when the foot leaves the ground (toe off). At this point knee flexes 35 degrees and it represents the second highest axial force point in the cycle, after this point the force is exponentially decreasing.

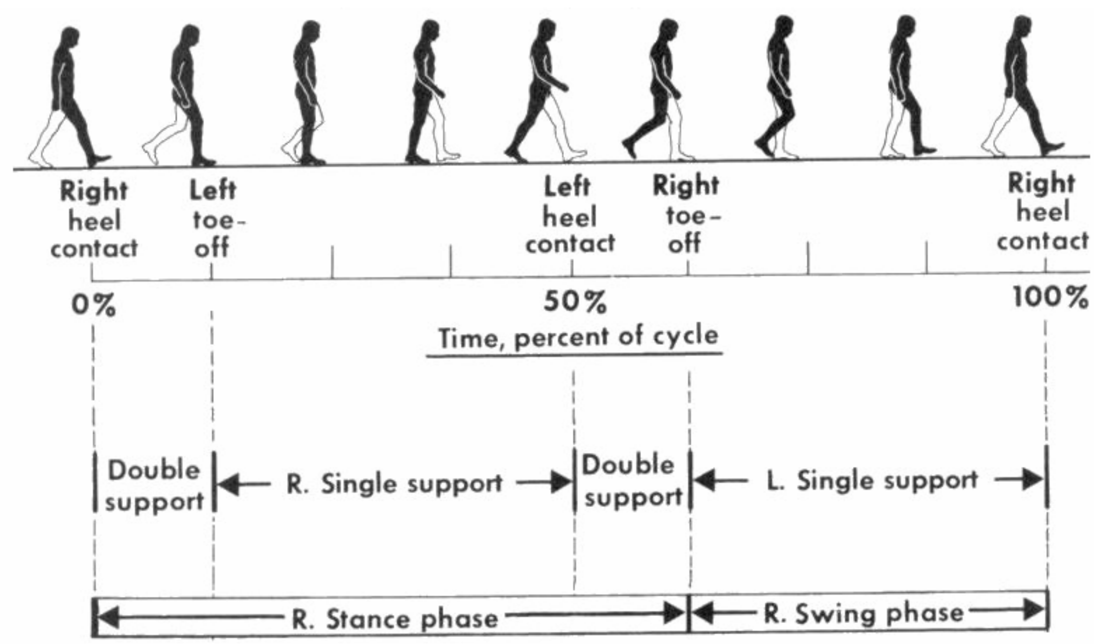

Fig 4.3 Distance and time dimensions of gait cycle

Swing phase represents 40% of the cycle and in this phase forces are mostly applied in the contralateral foot. This phase consists of 3 stages:

1. Initial swing: It occurs when the toe off until maximum knee deflection. At this point, the contraction of quadriceps initiated before toe off serves to prevent heel from rising too high in a posterior direction and also helps to initiate the forward swing.
2. Mid swing: This stage is from the maximum knee flexion until the tibia is perpendicular to the ground (without touching it).
3. Terminal swing: It begins when the tibia is vertical to the ground and ends at the initial contact. At this point, the hamstring muscles act to decelerate forward swing and control the position of the foot at heel strike.

In order to simulate a dynamic analysis in the knee joint, the three most significant parts of the cycle will be analyzed by separate, these stages will be:

1. Full foot strike, that is at the contralateral foot toe off,
2. Toe off, the exact point when the foot leaves the ground
3. Mid swing, just when the knee is at its maximum flexion point.

ISO standard also has a study of wear of knee joint, 14243 – 1.4 standard according to this standard will be obtained the axial force, rotation torque and anterior posterior force applied in knee joint during each stage of the gait cycle (Appendix B). These forces are shown in Table 4.2.

Table 4.1. ISO 14242 – 1.4 Wear of Knee Joint

		Full Foot Strike	Toe Off	Mid Swing
Angle	[^{o}]	15.145	8.283	58.022
Axial Force	[N]	2577.311	2416.77	170.514
Anterior-Posterior Force	[N]	112.210	-137.56	48.317
Torque	[N · m]	-0.887	5.740	0

4.4. First Draws

Drawings were made in Pro-E from a real knee model provided by an orthopedic doctor in Poza Rica, Veracruz. This model have the average measures of an American man, it was measured and then drawn by projecting different surfaces one over another, first a front view projection merged with the lateral view and finally a top and bottom view in order to obtain a real approximation of knee bones. To give more realism to the drawings and avoid unnecessary higher amount of stresses points and errors during the analysis, all or almost all the sharp edges were rounded. These drawings are shown in Appendix C.

4.5. Analysis

The analysis of this chapter, corresponded to analysis of the knee joint without biomedical implant was made in "mechanical" using the model previously drawn and transformed into a solid in Pro-E.

As we saw at the beginning of this chapter, some assumptions have to be made in order to perform a biomechanical analysis the assumptions made for each category are listed below:

- Geometry:

  The lozenge is not involved in the analysis because it is not in direct contact with bone as the femur and the tibia do. The lozenge is only attached to the knee joint by cartilage and muscle and is barely involved in supporting.

- Material:

  The corrosive environment of the human body is not taken into account; this analysis is made under normal conditions.

- Structure distribution:

  The bone is completely solid and non porous.

- Boundary conditions:

  i. The top part of the femur is fully constrained, it does not move. Only the tibia and fibula are free to move.

  ii. The union between femur and tibia is totally constrained in all moving axes, and only allowed to rotate in the Y axis. This meaning that the tibia can not go above the femur and than despite the fact that the human knee can rotate around 15 degrees in the x axis it is not taken into account.

The first step to realize the analysis is to ask the program to “mesh” the model, i.e. analyze the geometry and divide it into the different finite elements that will be analyzed by separate and which will show an individual result. The mesh was done of a medium size in order to have a good but fast approximation of the elements in the knee.

After the program did the mesh constrains where applied as we have discussed previously and shown in the figure 4.4.

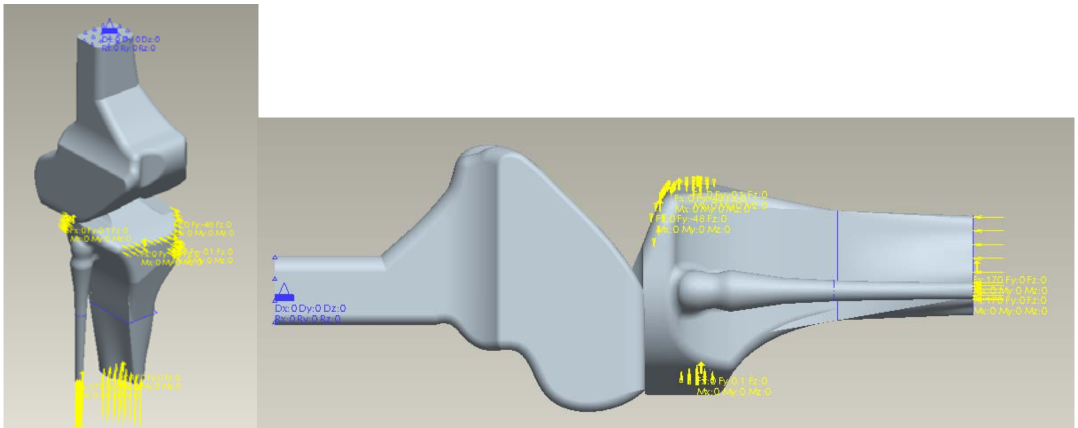

Fig 4.4. Application of constrains and forces during the analysis.

After applying constrains, forces were applied; the forces applied were discussed previously and were applied as shown in table 4.2. The method of application was obtained from the ISO 14242 – 1.4. The wear of Knee Joint standard is shown in figure 4.5.

Table 4.2. Axis and magnitude of the forces applied.

		Axis	Full Foot Strike	Axis	Toe Off	Axis	Mid Swing
Axial Force	***[N]***	X	2577.311	X	2416.77	X	170.514
Anterior-Posterior Force	***[N]***	-Z	112.210	-Z	-137.56	-Z	48.317
Torque	***[N · m]***	-	0.887	+	5.740		0

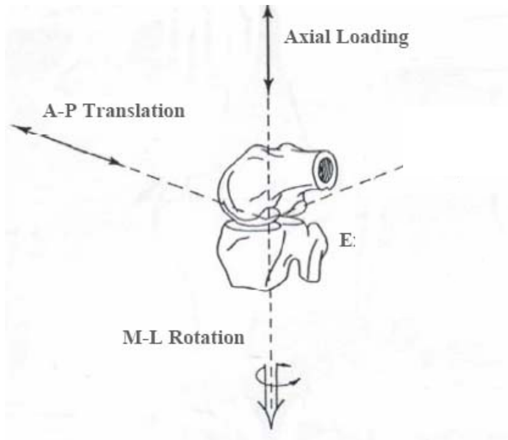

Fig. 4.5. Direction of the forces along the knee joint

Once all the external elements for the analysis were applied, the next step was to give to the structure the properties that the software needed to perform the stress and strain analysis. These properties where the Young's Modulus is 16 GPa, the Poisson ratio is 0.33 considering an isotropic bone and finally a coefficient of thermal expansion of $27.5x10^{-6}$ 1/°C. [43]

As a final step, the program makes the analysis, the conditions were a 12 color pallet, structural element, for stress and strain analysis and as the top part of the model is totally fixed, and the axial force is higher than the others in magnitude in the positive x axis this meant a compression force applied, the results asked to be shown were the minimum principals.

An analysis for each of the three principal stages of the gait cycle was performed and these results and their analysis are shown in Chapter VI and in appendix D.