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STRESS ANALYSIS LOADS ACTING IN THE HUMAN MENISCI HUMAN BY STIFFNESS METHODS

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Abstract. *The menisci are components of a knee joint, comprising two structures semilunar form, located within the joint capsule, lubricated by synovial fluid and composed of cartilaginous tissue.The meniscus, which, as stated, acts as a cushion receive these loads of magnitude of approximately twice the upper body weight, and, when at rest to redistribute; when moving beyond the redistribution, this damping will also thus enabling the body to move without further damage to the joint structure.*

This article aims to define the point of view biomechanical function of the knee joint, taking into account the degree of freedom of movement, analyzing the human meniscus as an isolated system, since the stresses acting aim to alleviate the burdens in the other system elements. By calculation whit the aid of MatLab program, we estimated the mechanical parameters after compression of the meniscus of knee flexion promoted by the starting upright up to four points of pre-set angle during movement of static squat without support,obeying the requisites provided in Solid Mechanics.

Keywords: : meniscus; biomechanics; knee; properties; mechanical stiffness.

1. INTRODUCTION

The condition of human motion is provided by an integrated system of bones, muscles and innervation: the musculoskeletal system. This system has a function beyond locomotion, sustain loads which the body undergoes both in movement and at rest.

The movement of a body happens through muscle contraction. The joints, in turn, have structures that function as pulleys, and load transmitting the motion to the other parts of the body. This load transmission determines the type, speed and positioning of the body in question.

The knee joint is an important part in the transmission of movement and load the lower limbs. Their interaction neuromuscular and skeletal system sets up a joint in hinge, with limited movement to a certain degree of freedom, transmitted by tendons and muscles that make up cables and menisci and articular capsule, which act as cushions distributing absorbing loads, besides the synovial fluid, doping the entire structure and also serves to absorb loads. The knee joints are often affected by injuries associated with sporting activities or everyday.

The investigations in biomechanics have been of fundamental importance both for injury prevention, as for therapeutic and discovery of new materials for the manufacture of prostheses and athletic shoes.

However, it has been difficult to obtain data for quantification of the charge which a joint is subjected. Therefore, it is essential to understand the mechanisms that contribute to the increase of these internal forces. The evaluation of internal loads are limitations in vivo, in that it becomes

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impossible to direct measurement without using invasive methods. According to type of movement, speed and angle of this, we have compression and shear forces acting directly on the joint, distributed along the contact areas between the structures that compose it.

The menisci are components of this joint, comprising two semilunar shaped structures located within the joint capsule, lubricated by synovial fluid and composed of cartilaginous tissue.We can make a clear analogy between some existing mechanical systems and the joints of the human body. The knee joint in this case represent a linkage composed of three segments (femur, tibia and fibula) mediated by a pivot pin (patella) for damped system cushions (menisci), with a lubrication liquid medium (synovial fluid) and movement and stabilization triggered by cable systems (muscles and tendons). This type of system, in this case, would function by moving a large structure with concentrated load above the center of gravity of the body, supporting and redistributing these charges so that the base of this structure can support its weight.

The meniscus, which, as already stated, acts as a cushion receive these loads approximate magnitude than twice the body weight, and while at rest to redistribute; when moving beyond the redistribution of this damping will also thus enabling the body to move without further damage to the joint structure.

The knee and composed of the tibiofemoral joint, tibiofibular, Patellofemoral. Has six degrees of freedom (movement angle).As the type of load imposed depends on the degrees of freedom, the areas where the loads will be concentrated over these types of menisci and also varies over the structure.

The present work aims to study the distribution of compression and shear loads acting on the meniscus during the squat to stand by via methods of determining stiffness, with the help of the MatLab program.

It is noteworthy that, as stated earlier, the menisci structure and biomechanics of the joint, in case it is treated as a mechanical system, not taking into account the biological characteristics such as chemical composition. In this paper we consider the menisci as a viscoelastic polymer biomechanical single and geometric features of uniform mechanical properties themselves, analyzing the meniscus as an isolated system, since the stresses acting aim to ease the load on the other system elements, and characterizing the forces acting on the meniscus in static squat movement, through the calculation of mechanical parameters (total voltage, system rigidity, deformation after compression deformation energy of the system) after compression and flexion of the knee squat movement promoted by static without support, via a program in MatLab.

The stress analysis would aim to determine the type of force acting on the biological material so well, the materials used can, besides the role to replace or protect the joint or bone, bring comfort to those who need its use, increasing the quality of life. Knowledge up the mechanical properties of the meniscus, we can determine or develop new materials suitable for use in biological prostheses and orthoses, and even in the development of products for sports shoes.

2- MATERIALS AND METHODS:

The objective of this study is to determine the distribution of the loads acting along the structure of the human meniscus , when an individual is at rest and in particular squat position . This is done from pre-existing data in the literature and the use of techniques of image manipulation and computer programs for mathematical resolutions (MatLab).

The application of the loads must always comply with the degrees of freedom of the joint , the knee angle determines the type of force applied to each body region of the meniscus , as these differentiate into compressive and shear according to the angle applied , may coexist within the structure as a whole or separately depending on the acting degree of freedom and magnitude of the applied force. Calculated parameters and the deformation of the meniscal body , respecting the principles of Solid Mechanics , calculate the deformation from an infinitesimal area determined by mapping graph to estimate the forces acting on the meniscal body , according to the area where each

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force acts , and the bending angle there of according to the joint position.

For this theoretical work, we considered the medial and lateral menisci as a single piece, without thickness variation, with the same properties and the same hardness throughout its structure (isotropic).For purposes of determining the total physical area of the human meniscus, held the graphic mapping through images provided by the literature.

2.1 - Balance Condition

To calculate the parameters proposed for the degree of bending required , is a program created by MatLab. To calculate the parameters proposed for the required degrees of flexion, a program is created by MatLab software, which is found in Figure 1.This program uses as input data the estimated weight in kg (later transformed by the same Newton) and linear anthropometric measurements of the individual in question, as the total height and measures of each segment.
For the procedure of data collection, the lateral and medial menisci were considered as a single piece, whose circular pieces representing the menisci have the same measurements of radii and single thickness with the same rigidity and properties throughout its structure (isotropic).

Each joint position, compression and shear forces acting on the meniscus, whose values depend on the degree of bending of this. In this paper, the joint positions chosen were 180 °, 135 °, 90 ° and 45 ° of knee flexion for representing the four positions where the compressive forces assume greater values and displacement related to the body of the meniscus in conjunction with the movement of other components of Articular capsule .

The meniscus human can withstand compressive forces of magnitude between 3.44 to 6.19 times the body weight and shear force directed posteriorly between 2.61 to 5.89 times the body weight, these forces act directly on the centers body mass and locations of each segment, as the body position has a direct influence on the position of these centroids. Considered the maximum force acting always, for all positions.

For all case studies, it is considered an individual weight 51 kg and height of 1.68 m. The action of this body are shown in Table 1. The measures were taken following standard protocol for anthropometric measurements used in fitness.

Table 1 - Linear measurements of the specimen by segment.

Segment	Length (cm)
Foot	13,00
Legs	38,00
thigh	42,00
trunk	55,00
Head (plus the neck)	25,00
arm	29,00
forearm	25,00
hand	17,00

Considered the initial length of the meniscus is 6.25 cm and its maximum deformation around 1.4% along its length axis linear cross. Therefore, the final size of the meniscus is 6.337 cm. The load weight acting on the power system will be related to the sum of the weights of the segments subtracting the weight of the legs and feet, as they are below the knee joint axis.

In a first step, the program calculates the segment and the total weight of the individual, taking as input the weight thereof. The program estimates through this given the weight of each element body and the sum of these gives the total weight of the body segments of the individual concerned.According.We can determine the weight per segment using anthropometric data in accordance with expressions shown in Table 2.

Table 2 - Body weight of the segments (Chandler et all, in:Investigation of inertial properties of the human Body.AEML Technical Report, PP 74-137.Wright –Pastterson Air force Base,1975)

Body Segment	Equations weight by segment
Foot	2. [(0,009 .WB)+2,48]
Legs	2. [(0,044 .WB)-1,75]
thigh	2.[(0,127 .WB)-14,8]
trunk	(0,532. WB)-6,93;
Head(plus the neck)	(0,032. WB)+18,7;
arm	2.[(0,022.WB)+4,76]
forearm	2.[(0,013.WB)+2,41]
hand	2.[(0,005.WB)+0.75);
Total weight of the segments(N)	Σweight of the segments (N)

Note: WB=Total Weight Body

After calculation of the total body weight and the weight per segment measurements are made linear length of each segment in question, according anthropometric standards, with reference to their alignment along the y-axis being represented as the height of the shaft inserted in this with respect to the x axis, characterizing the input data of the system concerned. The location of the center of mass of each segment is the segment position relative to an axis of Cartesian coordinates X and Y locations of each element of an individual, represented by the centroid position of the body in dependence on the angle position of the body segments in the axis y.

Table 3. Positional relationship between the angles of the segments relative to the y axis:

Segment	180º	135º	90º	45º
Foot	0	0	0	0
Legs	0	160	155	145
thigh	180	135	90	45
trunk	0	10	15	25
head (plus the neck)	0	10	10	15
arm	90	90	90	90
forearm	90	90	90	90
hand	90	90	90	90

The angles were considered as fixed in reference to the dynamic optimal exercise execution to obtain maximum yield during this, whose values are presented for each angle in radians / element.

The determination of the center of mass of each segment and overall, according to a coordinate axis X and Y locations of the medial and distal ends of each element, in relation to an

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individual static positioning of the biometrics is based on correlations, as described in equations 1,2,3,4 and 5:

$$\eta_{segment} = (\pi.\ \eta_{segment})/180 \tag{1}$$

$$X_{distal} = C_{segment}.\ \mathrm{Sin}\ (\eta_{segment}) + P \tag{2}$$

$$Y_{distal} = C_{segment}.\ \mathrm{Cos}\ (\eta_{segment}) \tag{3}$$

$$X_{proximal} = C_{segment}.\ (60/100).\ \mathrm{Sin}\ (\eta_{segment}) + X_{end} \tag{4}$$

$$Y_{proximal} = C_{segment}.60.\ \mathrm{Cos}\ (\eta_{segment}) \tag{5}$$

Where: ηsegment = angle in radians segment, P = anterior segment position on the x axis; Csegment =segment measurement in meters; Xdistal and Ydistal = distance between the track and the opposite end of the main body; Xproximal = distance from and following Yproximal nearest end with respect to the main body.

The location of the center of mass of each segment and overall, according to a coordinate axis X and Y locations (internal midpoints) each element of an individual static positioning relative to this obeys the following correlation according to the equations 6 and 7.

$$X = \Sigma\ [(P_{segment}.\ X_{proximal})\ /\ W_{ts}] \tag{6}$$

$$Y = \Sigma\ [(P_{segment}.\ Y_{proximal})\ /\ W_{ts}] \tag{7}$$

Where: $P_{segment}$ = weight of each segment; Wts = total weight of the segments.

2.2 - Determination of Strain

2.2.1 - Discretization Area

For purposes of determining the total physical area of the meniscus human mapping is performed by graphic images provided by the literature.With such images, it creates a graphical grid, using the Excel program and this was inserted as a backdrop, this is then divided into subareas regular grid, as shown in Figure 1.

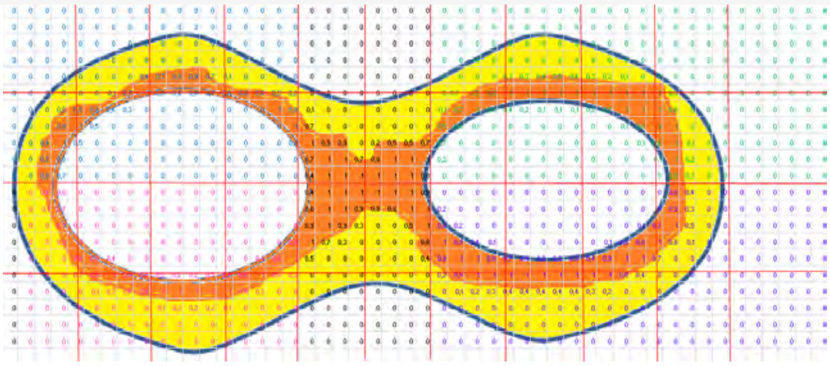

Figure1- Example mapping graph for joint position regarding the 90º.

It is noteworthy that these areas represented in each box, working independently of each other, like springs "bagged" of a mattress as shown in the example in figure 2 .

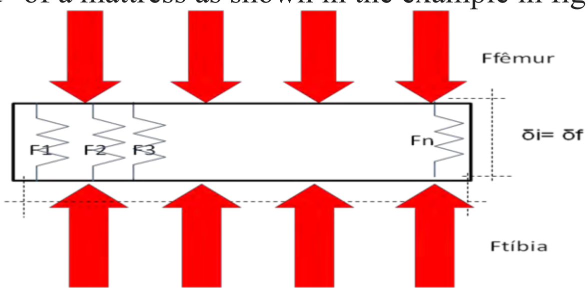

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Figure 2- representative example of the meniscus in the form of springs to the initial position of the knee joint.

After the creation of the grid for each squat position considered, related to the components of the system to which the meniscus inserted this (femur and tibia), it was marking the areas under compressive stress at each position of interest.The demarcation of areas of contact follow the movement of the structure, according to the joint position required, according to the scheme of the figures 3(a) and 3(b).

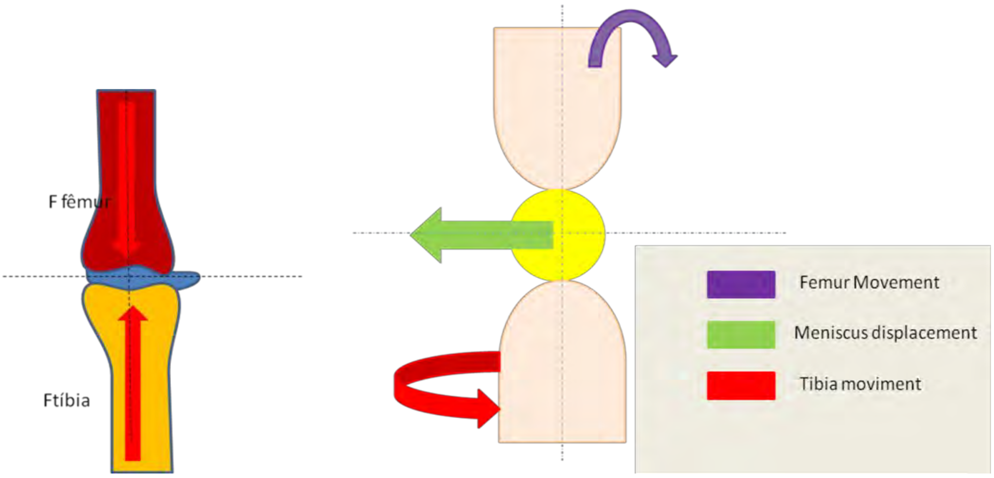

Figure 3: (a) Example of a diagram of forces acting on the joint at rest;
(b) Scheme of joint movement during knee joint flexion.

Then the mapping is done itself; fully inserted within the areas of grid cells are assigned the numerical value 1 and the other, when a value ranging from 0 to 1, according to the insertion of the marked area within the figure in relation to the total area represented. Grid cells is created measures 21.4 X 15 mm, with a total of 1,352 squares. The approximate measurements without distortion, the meniscus human correspond to approximately 3.1 cm to 3.15 cm medial meniscus and a lateral meniscus to, giving a total linear extent corresponding to about 6.25 cm.

By summation of the measures and dividing the total area of the figure by the total area of the grid, encontrare is a conversion factor for each grid, corresponding to 0.00884 .

This admensional conversion factor is required to adjust the parameters calculated at the actual area of the body in question, once the mapping is done in the figure larger than the size of the body concerned.

Each box represents a sub-area of the figure which corresponds to a spring, therefore the system has a total of 1,352 springs, acting alone or together, under compression in a certain joint position. The possible movements for the articulation of the knee are flexion, extension, rotation, and translation of the meniscal body in reference to the movement performed by the tibial plateau.

2.2.2 - Compression of the Meniscus

The equations governing the uniform compression of a body are:

$$W = \int_{At} \sigma \, dA \tag{7}$$

$$K = (A_t E) / L \tag{8}$$

$$\delta = (P.L) / (A_t E) \tag{9}$$

$$E_{elast} = ½ P^2 L / (A_t E) \tag{10}$$

Where: W = force applied to the system; A_t = total area of the system;σ = applied stress by area;P = compressive force applied to the system;δ = linear deformation (displacement) of the system after the action of compressive stress;K = total stiffness of the system;E = modulus of elasticity of the material;L = final size of the body after deformation;E_{elast}= strain energy of the material.

For the parameters relating to each particular area of the charter equations are modified to suit local conditions of each spring considered. Thus, the local forces acting on the body can be described by the equations 11 and 12:

$F_i = (K_i / \Sigma K_j) F_{tibia}$; (11)

given i, j = 1, n

$F_i = K_i \delta_i$ (12)

Where: n = total number of springs;i = index for the spring study;K_i = stiffness coefficient for each site spring;F_{tibia} = less force acting legs and feet;F_i = force acting on each spring;δ_i = linear deformation suffered for each part or spring inserted into the meniscus.

2.2.3 – Flexion.

As seen, the menisci, being anchored to the tibia move in conjunction with this movement of describing translation and rotation. In Figure 4, we observe the movement of the body under the action of compressive forces acting on a Cartesian plane.

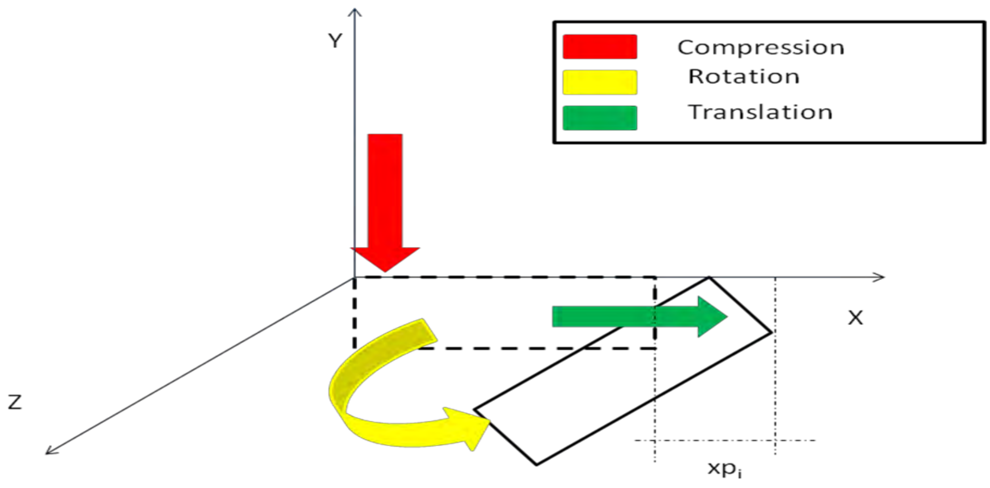

Figure 4 - Example of displacement of a body on a plane after applying compressive force at any point.

Considering now an angle θ of rotation of each spring undergoes a lowering given by:

$\delta_j = X_j . \theta$ (13)

$M_o = \Sigma (F_j X_j) = \Sigma (\delta_j K_j) X_j = \Sigma (K_j X_j^2).\theta$ (14)

Where: M_o = bending moment;δ_j = linear deformation undergone by each spring due to rotation of the assembly;X_i = the position (lever arm) of each spring with respect to the center of rotation.

Then, the angle of lowering the total system can be calculated:

$\theta = M_o / (\Sigma K_j X_j^2)$ (15)

For the spring force acting on each of an area P during rotation, has, a pressure differential:

$\theta = X_p$ (16)

$F_p = \theta. K_p.X_p$ (17)

Where: F_p = local force acting on the system; K_p = spring stiffness considered; X_p =spring position with respect to the center of rotation.

Thus, the equation for the force acting spring is:

$F_P = [(K_p.X_p) / (\Sigma K_j X_j^2)] M_o$ (18)

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The equations governing the translation are:

$$\Delta_i = \Sigma K_i (\delta_i) = \Sigma K_i .F \qquad (19)$$

F_{tibia} being the force acting on the system. Thus, for each area A_j in question:

$$F_j \delta = K_j K_j / \Sigma K_i F_{tibia} \qquad (20)$$

2.2.4 - Combined Compression and Bending

Once the model adopted in this work is linear, the action of the compressive force and the moment applied simultaneously meniscus is given by the sum of the expressions developed previously.

3-RESULTS:

After the initial calculations, in possession of the values of forces and moments acting, we can calculate parameters of the mechanical system: under tension for each areain accordance with the degree of flexion of the joint in question actuating tension (σ) ,lowering system (δ), and stiffness (K). The approximated maximum and minimum values on the properties studied and their variations corresponding to the the table below, respectively:

Table 4-Maximum and minimum values for stress, strain and stiffness, obtained for the joint positions 180 º, 135 º, 90 º and 45 º.

Joint Angle	Stress (σ.[N.m])		Strain (δ.m)		Stiffness (K)	
	σ_{max}	σ_{min}	δ_{max}	δ_{min}	K_{max}	K_{min}
180º	0	$(-5,38).10^{7}$	$(0,95).10^{-11}$	$(0,01).10^{-11}$	$(9,8).10^{4}$	0
135º	0	$(-5,70).10^{7}$	$(0,18).10^{-12}$	$(1,66).10^{-12}$	$(2,32).10^{4}$	0
90º	0	$(-6,62).10^{7}$	$(1,25).10^{-11}$	$(0,1).10^{-12}$	$(2,66).10^{4}$	0
45º	0	$(-0,5).10^{7}$	$(3,5).10^{-11}$	$(0,4).10^{-12}$	$(1,61).10^{4}$	0

At 180 ° of knee flexion, there is a flexion of the body of the meniscus, however insignificant for the purposes of axial stress relief;the compression only consider, in this case, acting almost uniformly.The system undergoes a slight rotation due to the stabilization the structure, and the compressive strength does not produce translation of the meniscus body; starting from 135, the tibia begins its rotation, causing the body of the meniscus to initiate a translatory movement being limited by the patellar body.

4-CONCLUSIONS:

Through this study we can observe the influence of the the position of articular contact Femur- Meniscus -Tibia relative to its position during flexion of the knee joint in the execution of static squat exercise. The calculations show that the highest surge tensions are at the start of the execution of the exercise, in any of the positions analyzed, which is considered to be critical phase of the same.

4.1 - Behavior of the body in relation to stress and strain for each joint position after application of force:

Regardless of the position in which the articulation is flexed, the body in question will have the same behavior in respect to the stress-strain curve.

The graphs show that the largest voltage peaks are at the end of the execution of the exercise, in any of the positions analyzed, which is considered a critical stage.

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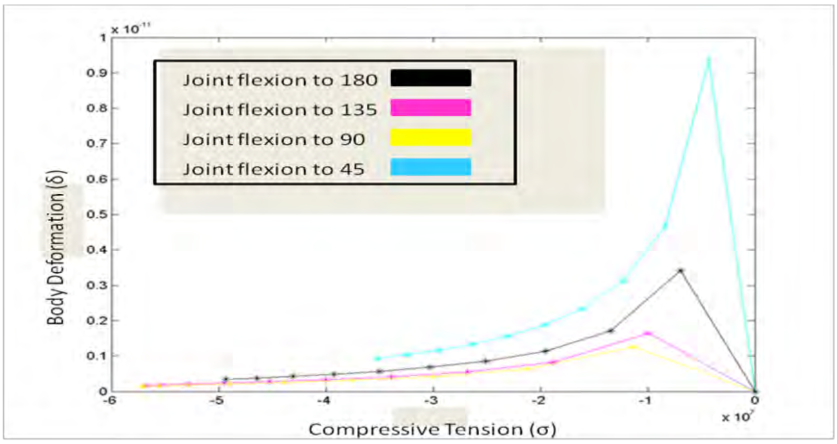

Fig.3-Comparative graphical representation tension x deformation for joint positions.

At the beginning of the movement, this degree of joint flexion, the stress suffers little peaks for the end of the rotational and translational movements. The lowering of higher degree in the lateral meniscus arises from tension generated by the accommodation of the femur bone snap in the structure / position.

4.2 - Behaviour in Relation to Body Stiffness and Deformation for each position after Articular Force Application:

With regard to rigidity, there is a greater value in the first third of movement, as this is resumed from a position above and the relevant part refers only to deformation by compression. The stiffness at 90 assumes the largest value for the start of the rotation and translation tibial which becomes a facilitating factor for these movements. The curve decreases during translation, caused by the increase of tension during accommodation anatomical structure, decreasing gradually until the end of the movement to be limited by the body of the patella, with decrease by the end of its accommodation complete.

The others positions show the same behavior analyzed, decreasing until the end of the movement, stiffness also became almost uniform after compression, since the tibial rotation is minimal and only for purposes of anatomical accommodation and there is no translation of the system.

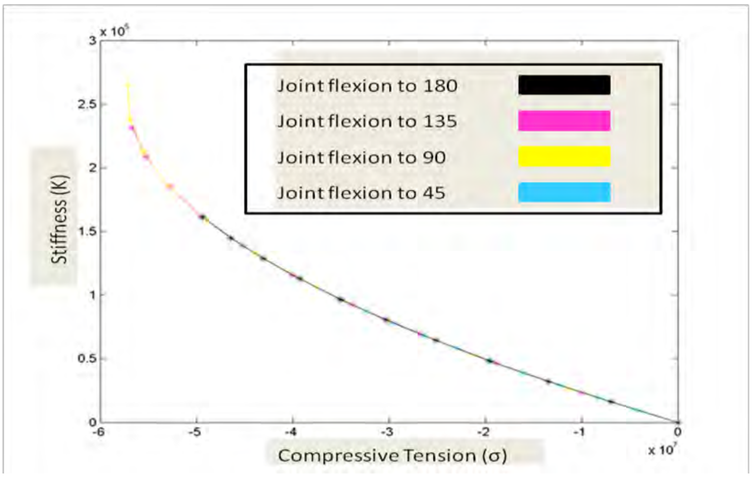

Figure 4 - Graphical analysis comparative general stiffness acting on the meniscal body.

In accordance with the observation of forces acting, this study has resulted in a set of concepts and results, aiming to characterize the behavior of the meniscal body under the action of a compressive force.

In overview, we observed that the articular position directly influences the stress and deformation of the meniscus, while the type of movement suffered by the given the body within the articular capsule inversely affects the material stiffness in question, taking into account the relative values for each spring inserted into the system.

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