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Biomechanics of human knee joint under dynamic conditions

Incorporation of numerical methods to analyze the biomechanical behaviour of a human knee joint is the conventional practice. The biological joint simulation studies require incorporating the material, and geometric non-linearities while developing a numerical model. In case of biological joint dynamic analysis, the preprocessing of Finite Element models will be a challenging job and needs huge computational requirements. As the biological tissues are highly non-linear, obviously the researchers face difficulties in handling material and geometric non-linearities. The aim of this study is to evaluate the contact mechanics behaviour of a human knee joint under various loading conditions. This study explores contact parameters in line with contact mechanics approach, which deals with frictional stresses at the contact interfaces of the knee joint, identifying the failure prone zones in the corresponding soft tissues, and modal response of the knee joint. This study provides the biomechanical characteristics of a human knee joint contact interactions that can be used as a surrogate models in complex dynamic simulations.

Keywords: Non-linear stiffness; knee biomechanics; frictional stresses; natural frequency; contact mechanics.

1.0 Introduction

The human knee joint is a very important part of the human body which allows the body to do very essential tasks like walking, running, standing, sitting, etc. The knee joint consists of different tissues made up of non-linear materials which function together. The different tissues of the knee joint are shown in Fig.1. The knee joint is a hinge/pivoted type synovial joint which joins the thigh bone called femur to the shin bone called tibia.

Ligaments: They provide stability to the knee. The ACL or Anterior Cruciate Ligament and the PCL or Posterior Cruciate Ligament prevent front and back movement of the femur relative to the tibia. The MCL or Medial Collateral Ligament and LCL or Lateral Collateral Ligament prevents the lateral or sideways movement of the femur. Ligaments are described as highly non-linear soft tissues.

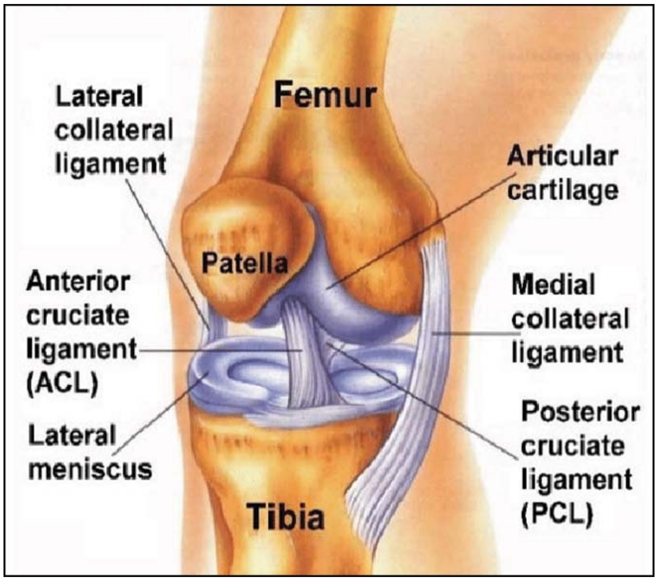

Fig.1: Anatomy of human knee

Menisci and Cartilage: The two C-shaped pieces of cartilage, Lateral Meniscus and Medial Meniscus are the shock absorbers between the femur and tibia. The menisci are made of fibrocartilage which is strong and rubbery in nature. Whereas, the end of the femur and tibia are covered with the articular cartilage also known as hyaline cartilage which is flexible and slippery in nature. Cartilage and menisci are defined as soft tissues because of their material behaviour.

Femur and Tibia: These are the two bones which are connected by the knee joint. Femur is the thigh bone and tibia is the shin bone. These bony tissues are said to be rigid in nature and highly stiffer in nature compared to soft tissues of the knee.

2.0 Literature survey

The menisci tissues found within the knee-joint and articular cartilages are attached on the end of the long bones. These two soft-tissues are functional in providing low-friction surfaces and supports in bearing the load to facilitate the function of the knee-joint (J Buckwalter and H Mankin, 1997).

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The Osteoarthritis (OA) and excessive soft tissue deformation are the primary diseases which are leading cost of conditions across the worldwide. In 2019 the global knee replacement market size was costing around USD 9.45 billion and is predicted to be reach USD 12.48 billion on 2027. The knee arthroplasty surgery is the most common surgery that replaces the damaged knee tissues by artificial knee implants. The tearing of meniscus is the common injuries of the knee which needs the immediate clinical consultation to avoid the degenerative risks (Aagaard, and Verdonk R, 1999) (Englund et al., 2003) (Stensrudet al., 2014). Menisci and cartilage tissue repair is widely used treatment for knee joint injury (James et al., 2014). Though the complete meniscal repair is not possible but partial repair could be possible by meniscal allograft to prevent the sequence of degenerative diseases (Yoon et al., 2014) (Vundelinckx et al., 2014). The measure demerit of the allograft treatment is the limited availability of donor tissues (Tugwell et al., 2005). As of today, the developed synthetic or natural meniscal implant is ideally not performing similar to the functionalities of fibro-cartilaginous tissues of natural menisci (Marsano et al., 2006) (Kon et al., 2008) (Kon et al., 2012). The decellularised natural xenogeneic tissue for meniscal replacement is the potential treatment for the mentioned diseases (Stapleton et al., 2008). The natural xenogeneic tissues could be supplied as class 3 medical devices, and there are no such problems associated with donors. It is highly desirable to preserve the biomechanical properties of this ‘off the shelf’ natural xenogeneic tissue implant to assure the proper functioning of the meniscal tissues of the knee. Also, off-the-shelf xenogeneic tissue could be available with any size, geometry, and flexibility in supply.

The current work’s primary objective was to investigate the stress-stiffening and biomechanical behaviour of meniscus and articular cartilages of the knee-joint through computational methods. The surrogate models developed from computational methods will ultimately assist in choosing the required properties for implants. To make this study as close to the real conditions as possible, the data and the boundary conditions were acquired from actual studies on human knee joint.

3.0 Material properties

The selection of the material characteristics to better reflect tissue behaviour is one of the challenging steps in biomechanical studies since the material properties of hard and soft tissues remain somewhat controversial and might vary between in vivo and in vitro (Kluess et al., 2009). That explains the difficulty of testing tissues’ material properties in vivo, leading to such controversy in assigning appropriate material properties in analytical studies. Many knee joint analyses studies have investigated the role of the bones’ mechanical properties, the articular cartilage, and the menisci (Zheng et al., 2014) (Zielinska et al., 2006) (Adouni et al., 2012).

Whenever the long loading times during the stance phases of the gait are involved with a normal viscoelastic time constant, then the non-linear material models shall be implemented (Pena et al., 2006) (Haut et al., 2002). (Venkatesha B K et al. 2014) studied the numerical analysis of damage tolerance design. This study focuses on linear material models and non-linear material models will be studied in next phase of this project. Constitutive models in ANSYS are applied to define material behaviour. The material properties which are reported in the past works were collected (Yang et al., 2010) (Guo et al., 2009).

Table 1: Material properties

Tissues	young’s modulus (mpa)	poisson’s ratio	density (kg/m^3)
Bone	20000	0.3	1830
Meniscus	59	0.9	800
Cartilage	12	0.475	800

4.0 Methodology

The 3D model of the knee joint was resourced from the open-source research repository (SimTK.org). The model is a cadaver data of a 53-year-old man having an early phase of osteoarthritis disease. This model was used for the analysis in ANSYS. Different types of analysis that are conducted on the knee joint are static structural analysis, modal analysis and harmonic analysis. The results obtained from these analyses were used to interpret the biomechanical behaviour of the knee joint the results of which can be used for further study or design of artificial knee joint or a prosthetic leg.

4.1 Pre-processing

In this analysis, the weight of the person that would be acting on the joint was taking into consideration and a vertical load of 708 N was applied in the negative Y direction. Since the model was obtained from the cadaver data of a 53-year-old man, the boundary conditions and loading were decided keeping in mind the weight and the physical conditions of that man.

4.1.1 CAD model preparation and establishing

The 3D CAD of the knee is imported in the IGES file format into ANSYS design modeller platform. The geometrical clean-up is carried in order to eliminate the redundant features of the CAD model. As the joint kinematics is not under the scope of this study, hence the fibula bone, patella bone, patellar tendon and patellar cartilage are removed.

4.1.2 Contact definitions

The penalty-based contacts are defined between different mating tissues in the model. Hence this requires parts to remain in contact with each other for proper force distribution and avoiding material penetration. Considering these parameters, the contacts were defined as given in Table 2.

TABLE 2: CONTACT DEFINITIONS

Interfacing tissues	Contact type
Distal femur - femoral cartilage	Bonded
Femoral cartilage - meniscus	Frictionless
Meniscus - tibial cartilage	Bonded
Tibial cartilage - proximal tibia	Bonded

4.1.3 Meshing

The articular cartilages are fine meshed with the tetrahedral elements of global mesh seeding size of 1mm. The geometrical regularity of menisci will allow fine meshing with hexahedral/brick elements with global mesh seeding size of 1mm. In this analysis, as the bone components are of least focus, and meshed with course mesh through tetrahedral elements of global mesh seeding size of 5 mm in order to reduce the computational requirements. The mesh parameters listed in the Table 3.

TABLE 3: FREQUENCIES AND MODE SHAPES

Mode number	Frequency in hz	Type of deformation mode
1	63.742	Bending
2	118.09	Warping, elongation
3	181.22	Torsion about Y-axis
4	258.88	Femoral oscillation
5	331.54	Warping, femoral oscillation
6	345.74	Warping, compression

The deformation and stress distribution could be possible by hexahedral mesh because the tetrahedral elements tend to be stiffer than the hexahedral elements. Whereas for cartilages, the geometry is too complex and tetrahedral elements can better fit complex geometry. Since the elements are quadratic, the tetrahedral mesh would be more suitable for the complex geometry and the accuracy would almost be close to that of hexahedral elements. The 3D model with the mesh generated is shown in the Fig.2.

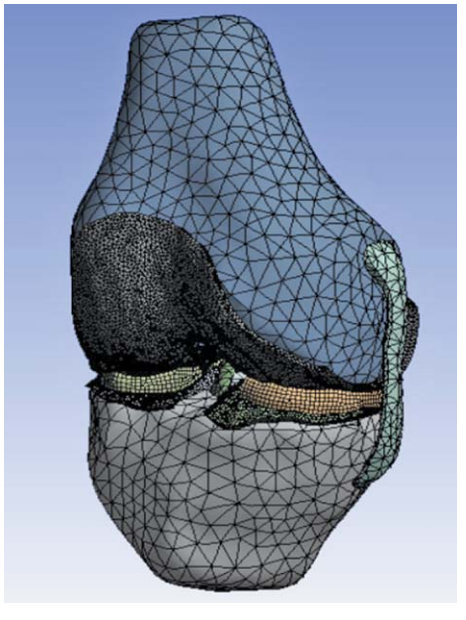

Fig.2: FE model of the knee joint

4.1.4 Boundary conditions

The vertical load is applied on the proximal surface of the femur through geometric features. The distal tibia surface is constrained to ground through fixed support. In this analysis, any displacement other than in the vertical direction was undesirable. The coupled constraints of the femur and tibia are made by the all four ligaments to restrict the displacement in only vertical direction (y-direction), thus this makes sure the displacements in x and z direction are restricted. Since the analysis involves geometric non-linearity, the large deflection was allowed.

4.2 PROCESSING

The FE model is solved in ANSYS Workbench 19.2 on dedicated workstation with 32 GB RAM and 12 GB graphics card support. Approximately 2 hours of computational time is required to solve the mathematical equations through matrix calculation. A three different analyses were carried out; static structural, modal and harmonic response analysis.

4.2.1 Static structural

This analysis is in order to determine the biomechanical behaviour of the knee joint under the static loading conditions. The static response is identified by maximum compressive stress, equivalent stress and directional deformation contour plots. The contact interaction results are interpreted through contact pressure and contact stress in the light of osteoarthritis.

4.2.2 Modal analysis

The dynamic analysis is carried out to determine the natural frequency of the given knee joint system. Firstly, the model is solved for free-free run without any boundary conditions in order to check the mesh connectivity and to identify the fundamental mode shapes along with corresponding natural frequencies. As the human knee joint is subjected to various forces and frequencies, this analysis would also help in determining the dynamic response of the knee model to external excitation at various frequencies. Along with laying a foundation to further dynamic analysis, this data would also be helpful in designing an artificial knee joint or a prosthetic leg.

4.3 POST-PROCESSING

The analysis is independently solved for static and dynamic analysis. The von-mises equivalent stress, maximum compression stress, directional deformation and contact frictional stress post processed to interpret the solutions in the form of contour plots. The fundamental frequencies or Eigen frequencies are also post-processed to a tabulated form for better interpretation.

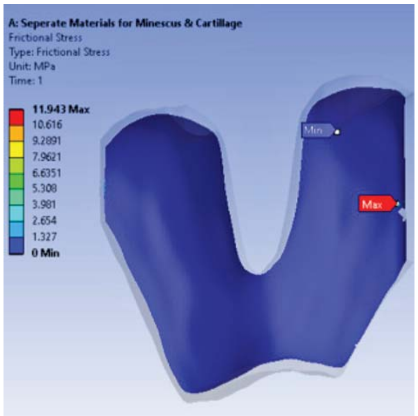

Fig.3 Frictional stresses in femoral cartilage

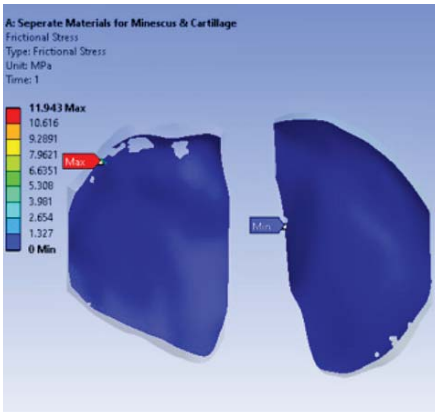

Fig.4 Frictional stresses in tibial cartilage

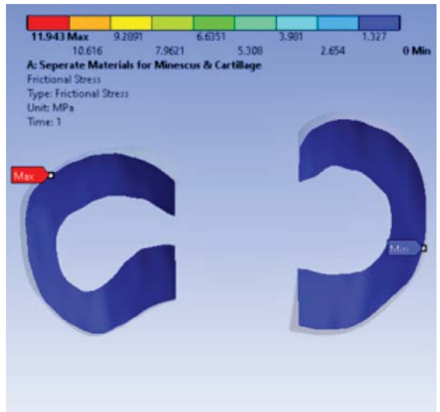

Fig.5 Frictional stresses in menisci

5.0 Results and discussion

The analysis of knee joint model is interpreted under the scope of contact mechanics analysis and modal analysis. The contact mechanics is studied under static structural context.

5.1 Contact mechanics

Fig.3 signifies the presence of frictional stresses in femoral cartilage with maximum value of 11.94MPa which is found to be exceeding the yield strength of the implant material. Even though the compression stresses are well below the tissue strength but frictional stresses are not in safer limits. This will lead to the beginning of the osteoarthritis at contact interfaces of the soft tissues, just because of the lack of lubrication and existence of the frictional traces. The significance of maximum frictional stresses in tibial cartilage and menisci tissues are found to be same as that of femoral cartilage, which are shown in Figs.4 and 5. It is noticed from these frictional stresses contour plots that the maximum and minimum values are concentrated on a comparatively very small region. Here the actual theme of this study lies in connecting the tissue degradation also because of frictional stresses.

The geometric non-linearity involved in this analysis can be explained by the force versus displacement curve given in Fig.6. This curve is also useful in describing the stiffness of the given joint. The non-linear stiffness of the given knee model again recalls the complexities involved in selecting material properties for implant design.

5.2 Modal analysis

Modal analysis is the first and very essential step of dynamic analysis. Firstly, the model is analysed for fundamental modes without any constraints, a free-free run, in order to check for the first six mode shapes and modal frequencies. When a modal analysis is done in a free-free system, the first six modes will be having zero frequency with rigid body modes resulting in addition to the elastic modes. These modes show the free translation and rotation of the system in six directions of motion and will be extracted as modes from one to six. Hence, a modal analysis is run on a free-free system and the frequencies of the first six modes are found to be zero.

The main aim of the modal analysis is to know the dynamic response of the model. The resulting data can be very useful in further harmonic analysis and also in designing artificial knee implants or prosthetic knee tissues. Once the natural frequencies are known, further study or designing can be done to avoid the dynamic failure because of resonance that can be catastrophic sometimes. The first six modes and their frequencies are given in the Table 3.

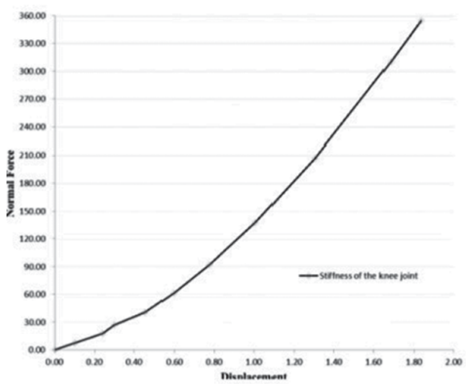

Fig.6 non-linear stiffness of the knee joint under static loading

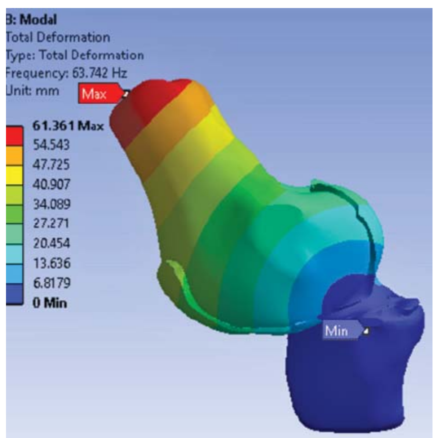

Fig.7: Eigen mode shape 1

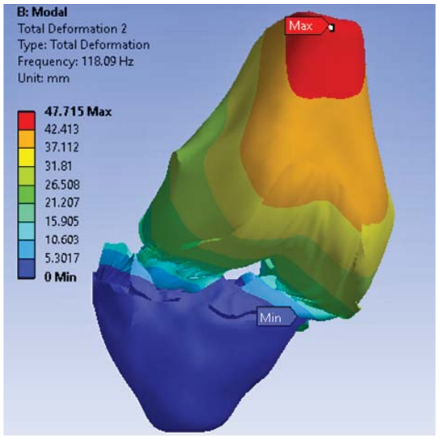

Fig.8: Eigen mode shape 2

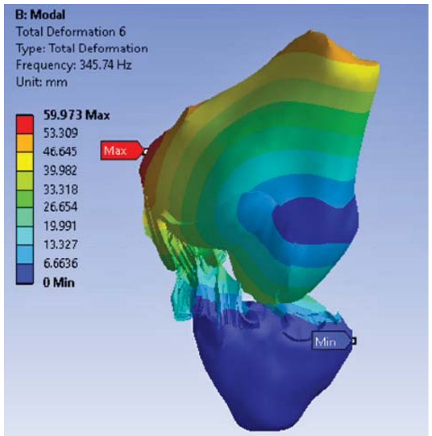

Fig.9: Eigen mode shape 5

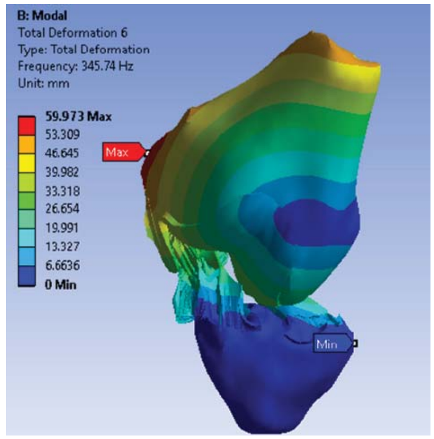

Fig.10: Eigen mode shape 6

Fig.7 shows the bending mode of deformation at an applied excitation frequency of 63.74Hzs. The warping mode of deformation is identified in Fig.8 wherein that occurs during 118.09Hzs. Fig.9 signifies the torsional mode of deformation which is found to be occurring at 181.2Hzs. The oscillation mode of deformation is recognized in Fig.10 that is found at 258.88Hzs. The Figs.11 and 12 are found to be combined mode of deformation which is described in the Table 3.

6.0 Conclusions

The health of a living joint generally depends on the material properties of the tissues constituting the joint and contact mechanics. This study explores contact parameters in line with contact mechanics approach, which deals with frictional stresses at the contact interfaces of the knee joint, identifying the failure prone zones in the corresponding soft tissues, and modal response of the knee joint. This study provides the biomechanical characteristics of a human knee joint contact interactions that can be used as a surrogate models in complex dynamic simulations. This study signifies the presence of frictional stresses at the contact interfaces with maximum value of 11.94MPa which is found to be exceeding the yield strength of the implant material. Even though the compression stresses are well below the tissue strength but frictional stresses are not in safer limits.

The stiffness formulations shall be used as surrogate joint characteristics in dynamic joint analysis using multibody dynamics and co-simulation.

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