MECHANICAL PROPERTIES OF HUMAN ARTICULAR CARTILAGE DETERMATION

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Abstract: The article deals with the mathematical modeling of biological tissues, the methodology and realization of experimental measurement of human hyaline cartilage is documented and carried out in vitro. The material characteristics of cartilage are presented on the basis of statistical analysis. The transversal isotropic material was chosen for its mathematical characterization and the constants Ei, Gij, µij (i, j = 1, 2, 3) were determined in compliance matrix *Sij*. These results were used for the FEM sub model of cartilage that was subsequently integrated into the entire model of human hip joint.

Introduction

Understanding causes of arthritic disease beginning also depends on exact description and clear definition of mechanical properties of human articular cartilage. Articular cartilage is nourished with the synovial fluid which is "sucked in" at unloading and "driven out" at loading. When this physiological situation is disrupted by a change of loading mode, a articular cartilage get damaged. An appropriate change of human locomotion during gait could possibly reduce the osteoarthritis development, which could be achieved by a special rehabilitation treatment. This work is one of several parts of major project. This project's goal is to define influence of mechanical loading on osteoarthritis development in human hip joint. Our advance has three steps: At first doctor determines degree of patient's disease by means of X-ray. In case rehabilitation therapy is suggested our project helps to improve that rehabilitation procedure. At second, rehabilitation worker monitors degree of patient locomotion by means of camera system and pressure-sensitive measuring board. Kinematics analysis is followed by finite element (FE) method analysis of human hip joint. FE analysis of arthritic and healthy hip joint was made in our laboratory. In the picture – Figure 1 you can see geometric 3D model and FE model of hip joint. It was created by means of set of CT cuts. The cuts were taken every 5 mm. In each 2D cut the curve was created around for example femoral bone. After that the obtained curves were turned into complete homogeneous surface. The whole model was meshed using the brick and shell elements by TrueGrid® program. FE model was made with help of Abaqus 6.5 software. Bones were modeled as rigid body, muscles as connectors and cartilage as deformable body. We are most interested in cartilage model. Shell deformable element was used for this. The goal of this project part was finding relevant material model of cartilage, working out of the appropriate methodology of material parameters determination, preparation and performance of experimental measurements.

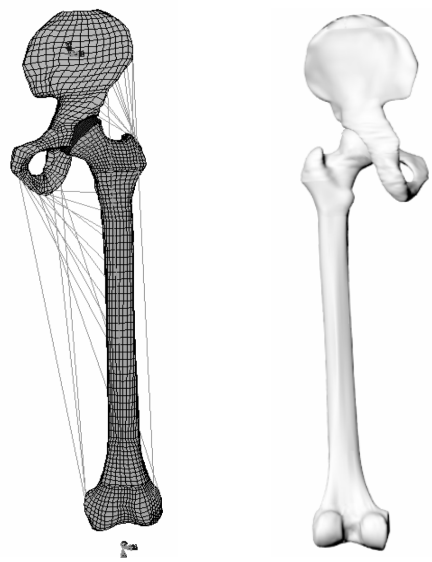

Figure 1: Finite element analysis and geometric model of human hip joint.

Articular cartilage is biological material which covers joint surfaces of articulating bones. In our project we obviously dealt with femoral and pelvic bones. Gaining of measuring samples was very difficult in our cases and it limited us. General problem for gaining samples is ethics and law – it could be topic for another discussion. The preparation, of course, had to be performed very carefully. For this reason, we prepared maximum of 15 samples from one femoral head. Cartilage consists of chondrocytes – cells of cartilage which shares only about 5 % of cartilage volume, collagens fibers which make the cartilage surface and matrix intercellularis [8, 9]. Cartilage is ingrown to the bone. Cartilage is fed with synovial liquid. According to medical literature cartilage consists of several layers. But nevertheless we did not succeed in identification of

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separate layers. That is why, we are looking at cartilage as a “black box”. We are not “concerned” in inner processes in cartilage, but we observe only its outer reaction to loading. On base of basic experimental measurements we found that cartilage has properties of orthotropic material. For this reason transversal isotropy material was chosen at the best for the cartilage model. For this model was necessary to determine four independent material constants.
The first part in this project was the experimental measurement and the material modelling of biological tissues. Transversal isotropic material was chosen for the human articular cartilage model and matrix elements for ***Sij*** were determined. Results of this work were applied in finite element (FE) analysis of hip joint [12, 13]. The aim of FE model together with kinematics analysis of patient motion is improved rehabilitation cure of osteoarthritis patients.

Materials and Methods

Forensic surgeon supplied resected femoral head. He marked points of orientation system. Only donors with no visible damage of joint were chosen. The specimens for measuring were prepared from the resected caput femoris, see Figures 2, 3. They were cut from the femoral head parallel and across to connecting lines (orientation system).
The donor was a man (age 21, weigth 75 kg, height 180 cm). The specimens had a block shape with these proportions – (2 x 2.5 x 5) mm. They were stored in physiological solution at temperature 5 °C. Before and during the measuring they were preserved in physiological solution at the room teperature (23 °C).

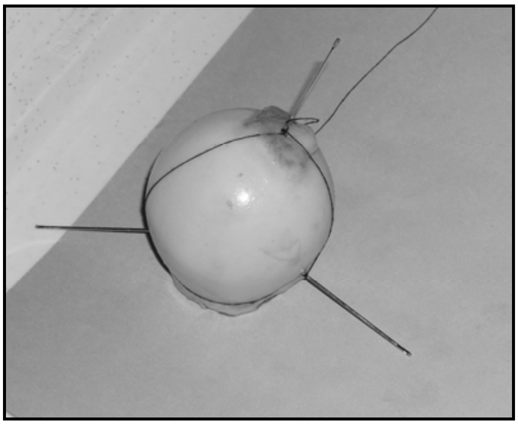

Figure 2: The resected caput femoris with the orientation system.

The specimens were loaded under the stress ranging between (1–9) MPa. [3] The orientation system was made by lines between *trochanter major – fovea capitis femoris – crista intertrochanterica.* Femoral head was placed into physiological solution at 5 °C before the beginning of experimental measurement.

The mathematical model of transversal isotropic material based on these mechanical tests was performed. These experimantal results were installed into the complience matrix ***Sij***, see Figure 4.

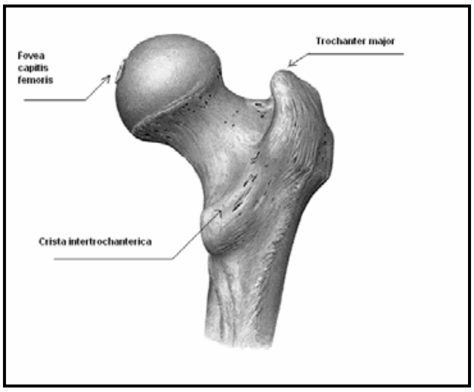

Figure 3: Proximal femur with orientation points.

The measurements have been carried out in Laboratory of Biomechanics on the Faculty of Mechanical Engineering of the CTU in Prague. We used test device MTS Mini Bionix 858.2 with special dynamometer for very little loading force volumes, see Figure 6. The stillness box was installed into this dynamometer. For surface morphology research we used stereomicroscope Nikon SMZ 1500 and pictures from this microscope were recorded by a digital camera. Measured data were saved at frequency 10 Hz. Loading and unloading velocity was 5 mm per second. Chosen loading range was (1–9) MPa. By this we simulated the degree of loading while walking, running and hard strike. The measured values were elaborated with statistical analysis using programme Statgraphics Plus 3.1 – you can see at the Figure 7, there is a example of Weibull distribution for elastic modulus in normal plane to the plane of isotropy.

$$\boldsymbol{Sij} = \begin{bmatrix} 1/E_1 & -\mu_{21}/E_2 & -\mu_{31}/E_3 & 0 & 0 & 0 \\ -\mu_{21}/E_2 & 1/E_2 & -\mu_{23}/E_3 & 0 & 0 & 0 \\ -\mu_{31}/E_1 & -\mu_{32}/E_2 & 1/E_3 & 0 & 0 & 0 \\ 0 & 0 & 0 & 1/G_{23} & 0 & 0 \\ 0 & 0 & 0 & 0 & 1/G_{13} & 0 \\ 0 & 0 & 0 & 0 & 0 & 1/G_{12} \end{bmatrix}$$

Figure 4: The transversal isotropic material compliance matrix.

The Young modulus values E were determined for each direction according to the hypothesis of equality verification ($E_1 = E_2 = E_3$). The resulting values were determined as the arithmetical mean values in each direction, namely E_1 = 144 MPa, E_2 = 201 MPa, E_3 = 217 MPa. The same principle was used for Poisson’s ratio determination and the modulus of elasticity in shear was calculated based on equations.

- $G_{23} = E_2/2(1+ \mu_{23}) = E_2/2(1+ \mu_{32})$ (1.1)
- $G_{12} = E_1/2(1+ \mu_{12}) = E_1/2(1+ \mu_{13})$ (1.2)

E1 is elastic modulus in normal plane to the plane of isotropy.
E2 = E1 elastic modulus in the plane of isotropy.
G12 = G13 is modulus of elasticity in shear in normal plane to the plane of isotropy.
G23 = G32 modulus of elasticity in shear in the plane of isotropy.
μ12 = μ13 is Poisson's ratio which reports transverse strain in isotropy plane to normal plane.
μ23 = μ32 is Poisson's ratio is transverse strain in isotropy plane [4, 5].
These material constants were used for FEM sub model of articular cartilage in hip joint [10, 11].
The choice of mathematical model was also influenced by experimental measurement results. In our experiment - each measured sample was loaded and unloaded by loading force in several cycles and the liquid was pressed out. It is seen clearly; see Figure 5, that high deformation of cartilage had place in first loading cycle. It was possibly caused by high content of water and forcing out of water from cartilage substance. During the following cycle water siphon age did not recur. The ultimate stability of cartilage behavior was reached during the gradual loading by seventh cycle. It is dare to say that water volume remained unchanged. We evaluated material parameters of TIM from last cycle. We replaced loading curve with straight linear relation with certain simplification. Tangent of straight line was searched value. We determined Poisson ratio so that we stopped at the certain loading volume and we measured transverse size of cartilage.

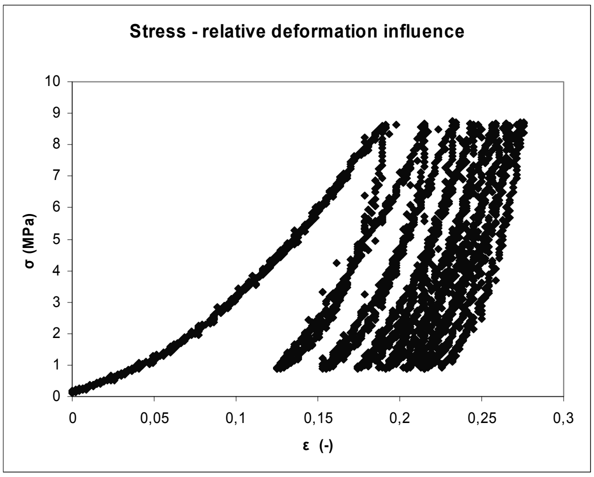

Figure 5: Loading cycles.

Great deformation will not occur in first loading cycle in reality. Pelvic-femoral joint is influenced by muscles groups even for unloaded joint. Cartilage is constantly loaded. Measured samples had surface layer cut. These sections absorbed water with less water resistance than surface layer. Conclusion is that the cartilage is loaded by muscles and another environment permanently for every moment of time. Synovial liquid is sucked and forced out in real life at fewer rates than during an experiment. Cartilage together with synovial liquid is lubricating system. Unlike it is sometimes said in literature – shock absorber. Reality is that cartilage began moving in the band of small deformation. Thus we could refer to theory of Hooks law.

Results

Table 1: Experimental parameters for cartilage model

E_1	E_2	E_3	G_{12}	G_{32}	μ_{32}	μ_{13}
[MPa]	[MPa]	[MPa]	[MPa]	[MPa]	[-]	[-]
144	209	209	56	83	0,26	0,28

Failure criterion will be determined by further experimental measurements, for the present σ_p = 28 MPa is used [1]. Contact cartilage–cartilage was modeled with help of cartilage friction coefficient which was taken from literature (f = 0,005). Shear stress which is necessary for cartilage-bone contact destruction was chosen as failure criterion, the phenomenon known at sport injures.
The results of this project should be used for identification of the significant causes of damaging and degenerative changes of the cartilage that leads to the immobilization of joint, verification and testing of bioimplants and development of new generation endoprotheses.

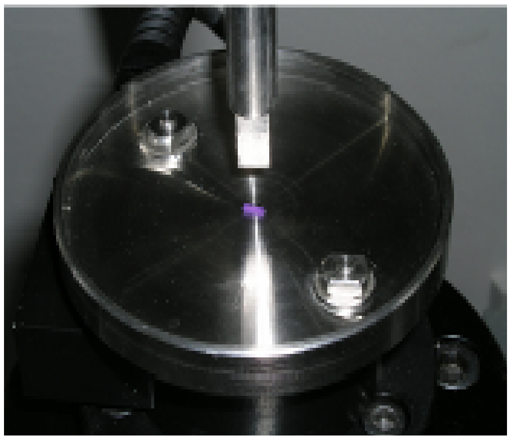

Figure 6: Cartilage sample in rostfest reservoir with physiological solution. The reservoir was installed into test device MTS 858.2 Mini Bionix

Conclusion

The articular cartilage, see Figure 8, 9, is a material inhomogeneous and anisotropic (orthotropic) [8, 9]. This biological material is composed of cells (chondrocytes), intercellular amorphous substance and fibrils [6, 7]. These components are very hardly separately measurable. Due to this fact it was necessary

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to consider this material as a finite element of continuum mechanics.
Monitoring the behavior of this material with the help of modern experimental method together with current computational software products could bring answer on causes of beginning of arthritic damage of joint.

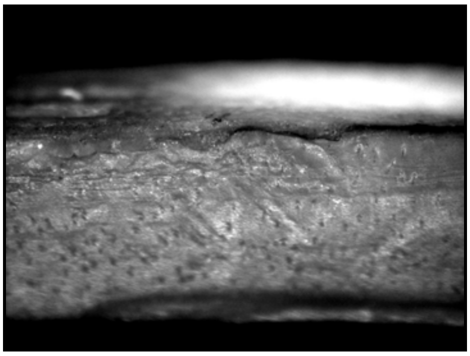

Figure 8: The cartilage in the longitudinal section, the surface is made of collagens fibres and inner layer contains the chondocytes (enlargement 180x).

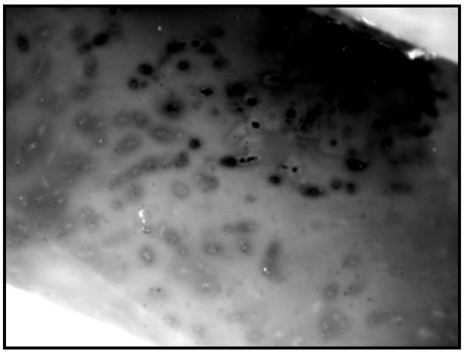

Figure 9: The cartilage surface with chondrocytes. This cut was ingrown with femoral bone.

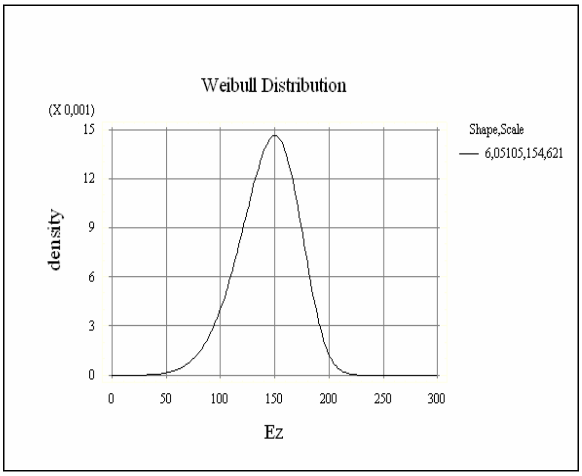

Figure 7: Weibull distribution for modulus elasticity E1.

Acknowledgements

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