

STRUCTURAL ANALYSIS OF ARTICULAR CARTILAGE OF THE HIP JOINT USING FINITE ELEMENT METHOD

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ABSTRACT

The paper presents the results of a preliminary study on the structural analysis of the hip joint, taking into account changes in the mechanical properties of the articular cartilage of the joint. Studies have been made due to the need to determine the tension distribution occurring in the cartilage of the human hip. These distribution are the starting point for designing custom made human hip prosthesis. Basic anatomy, biomechanical analysis of the hip joint and articular cartilage are introduced. The mechanical analysis of the hip joint model is conducted. Final results of analysis are presented. Main conclusions of the study are: the capability of absorbing loads by articular cartilage of the hip joint is preliminary determined as decreasing with increasing degenerations of the cartilage and with age of a patient. Without further information on changes of cartilage's mechanical parameters in time it is hard to determine the nature of relation between mentioned capability and these parameters.

Keywords: hip joint, pelvis, femur, cartilage, finite element method.

INTRODUCTION

For many years a phenomenon called population ageing, linked with continuing low or negative population growth rate, is being observed. The most tangible repercussion is growing demand for medical services dedicated for elder people.

Thanks to major breakthroughs and constant development on the field of medical imaging (mostly computed tomography and magnetic resonance imaging) it is possible to obtain more and more precise data on the anatomical structure of the hip joint. One of possible ways to make use of these informations is to perform a number of mechanical analyses of the hip joint and articular cartilage in particular, which then can be used as a base of assessment during the process of qualifying patients for hip replacement procedure.

Anatomy of the hip joint

A joint (or articulation) is a location of contact and movement of bones. A hip joint (Fig. 1) is a connection between femur and pelvis, with three degrees of freedom in three planes – coronal, median and axial [10]. Its articular surfaces have got the most regular shape compared to other joints in human body. The cortycoloid cavity (or acetabulum) is formed on the pelvis and covered with articular cartilage, but only in peripheral area [16]. Middle part, free of cartilage, creates the acetabular notch filled with fat tissue and synovial villi, whose purpose is to decrease pressure and friction between the head of the femur and acetabulum. The acetabular notch is a place of implantation of the ligament of the head of the femur [5, 11, 13, 16].

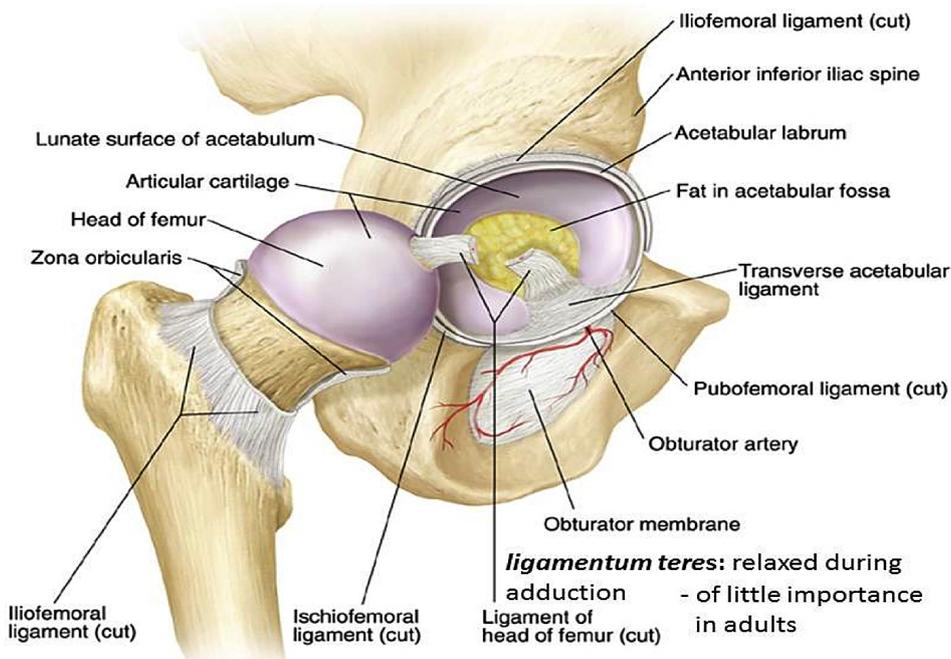


Fig. 1. Anatomical structure of the hip joint [18]

The head of the femur is globular with approximately 25 mm of radius for adult people. Practically all its surface is covered with articular cartilage, with the exception of the fovea capitis-femoris, which gives attachment to the ligament of head of femur [11, 13, 16].

Mechanics and loads on the hip joint

The hip joint is one of the most vulnerable to overstraining and degenerations structures in human. Its main role is to transfer body weight onto legs and enable rotational moves in all anatomical axis – horizontal, sagittal and vertical. These moves are: flexion and extension around a horizontal axis, lateral rotation and medial rotation around a vertical axis, and abduction and adduction around a sagittal axis [2, 16]. Thanks to the specific structure of bones and cartilages along with strong muscles and ligaments, the hip joint is capable of carrying high dynamic loads.

These loads influencing the hip joint are a complex system of forces and torques depending on body weight and structure, strength of active muscles, gravity and human activities. There are a lot of models of hip biomechanics, eg. Pauwels, Maquet, or Husikes [2, 16].

Structure of the articular cartilage

In a prenatal stage, human skeleton consists mostly of cartilage, while after birth its presence

is reduced to surfaces of joints, larynx, trachea, bronchi, nose and ears. Articular cartilage has an important role in the human musculoskeletal system. It is located on surfaces of joints, where most of the movement of the bones is executed. Cartilage transfers and spreads loads between bones, while preserving appropriate stress distribution on a surface of the joint, reduces friction, absorbs sudden overstrains and protects bones from surface wear. Fully developed cartilage does not contain blood vessels or nerves [3, 15, 16].

Articular cartilage has a layer structure (Fig. 2) and consists of 4 zones: superficial, middle, deep and calcified. In 75-80% it consists of water and in 20% of proteoglycans, collagens, glycolipids and chondrocytes forming cartilage matrix. Superficial zone, located in direct proximity of the articular cavity, is deprived of cells in favor of collagen fibers with the diameter between 5 and 100 nm, parallel to articular surface. This layer comprises 10% of cartilage thickness, and its main function is to protect structures below. In middle zone (40% of cartilage thickness) collagen fibers are diagonally oriented. Deep zone comprises 30-40% of cartilage thickness and collagen fibers are perpendicularly oriented. Calcified layer is a part of cartilage having direct contact with bone. Thanks to mixed orientation of collagen fibers articular cartilage maintains its proper mechanical properties, ensuring protection of a bone thanks to effective stress distribution and overstrain absorption. [3, 15, 16].

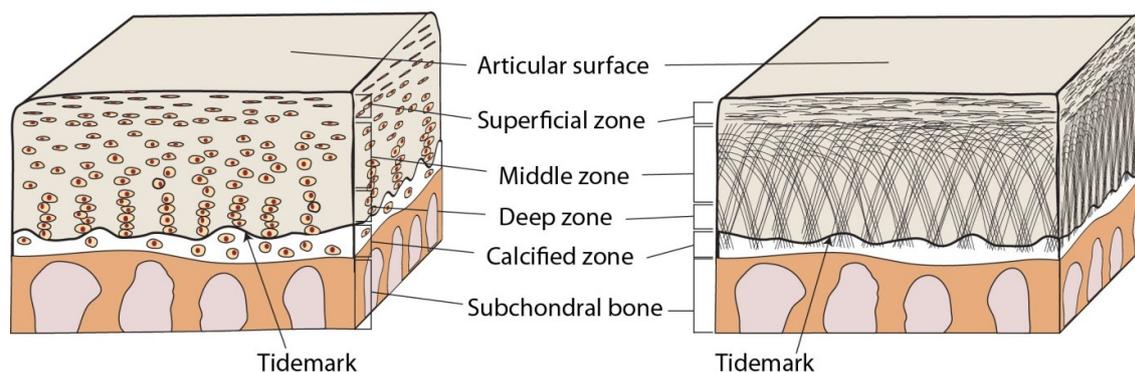


Fig. 2. Zonal features of human articular cartilage and subchondral bone [19]

Biomechanics of articular cartilage

The articular cartilage, while appearing smooth, is in fact a porous material. Its biomechanical properties depends on movement of fluids inside and outside the tissue, while the joint is loaded [16]. From the mechanics point of view the articular cartilage is treated as heterogeneous material with anisotropic mechanical properties. Fibers create a complex spatial structure, and the highest strength of the cartilage occurs in areas most affected with loads. Distinguished layers' differences aren't only in the direction of collagen fibers, but also in their concentration. In superficial layer of healthy cartilage concentration of fibers varies between 16% and 31% of tissue's volume, while in deep zone its average value is 14% to 42% [1, 3]. Estimated longitudinal strength of cartilage fibers on the head of the femur is 35 MN/m², whereas lateral strength stands at 18 MN/m². Damage of the cartilage may occur under influence of loads between 1 and 30 N/mm². The strength of subchondral bone is approximately 20 to 35 times larger than the strength of the cartilage. [1, 2, 4]

The surface layer exhibits the highest value of Young's modulus and also the minimum compressive strength [6, 7]. The intermediate layer is characterized by high compressive strength, the strength is 25 times higher compared to the superficial layer [6]. Through a combination of layers of varying structure is formed tissue to withstand large compressive and shear stress to protect the elements of the movement. The stiffness of cartilage evaluated using a summary module is in the range of from 0.5 to 0.9 MPa [6, 12]. At equilibrium, Young's modulus has a value from 0.45 to 0.8 MPa [6].

Damage caused to the articular cartilage

Disturbances in the metabolism and structure of articular cartilage are most often the result of an overload of the passage organ motion and its injuries. These include osteoarthritis and chondromalacia. The main causes leading to joint damage and changes in their mechanical performance is usually their mechanical overload lesions of rheumatoid ground, changing the properties of the synovial fluid, bone or joint deformity, local destruction of the articular cartilage [1, 15].

The term chondromalacia often identified with the so-called. dry the cartilage, it is a disease leading to the destruction of cartilage. Early pathological changes of chondromalacia include the limited field-drying cartilage and its swelling. As the disease progresses cartilage surfaces are separated from deeper layers. This results in the formation of cysts and ulcers and consequently local damage to the collagen fibers which contributes to the formation of fistulas and fragmentation of the cartilage. If the process continues may be completely expose the subchondral bone. If the joint surfaces of the part altered by disease processes cartilage structure and properties of the subchondral bone may be other mechanical properties [1, 15].

Software tools used in the study

The model used in the study was created by using specialized software. Firstly, multiple CT scans were used as a source for creating a base 3D model in Materialise Mimics. Thereafter, the obtained model was processed and modified in Solid Edge ST8 in order to approximate the articular surface of the hip joint. The conducted approximation was crucial to achieve more accurate analysis.

Materialise Mimics is software specially developed by Materialise for medical image processing. It is used for the segmentation of 3D medical images (coming from CT, MRI, 3D Ultrasound), resulting in highly accurate 3D models of patient’s anatomy. These patient-specific models can be implemented in a variety of engineering applications directly in Materialise Mimics or Materialise 3-matic, or exported to 3D models and anatomical landmark points to 3rd party software, like statistical, CAD, or FEA packages [17, 20].

Solid Edge is a 3D CAD, parametric feature and synchronous technology solid modeling software. It runs on Microsoft Windows and provides solid modeling, assembly modeling and 2D orthographic view functionality for mechanical designers. Through third party applications it has links to many other Product Lifecycle Management technologies.

Implementation of two highly-efficient graphic modelers – Parasolid and D-Cubed that allows combining direct modeling with precise control of geometry and gives engineers opportunity to conduct the designing process with speed and simplicity on a level, that has never been seen before [9].

STUDY ON STRESS DISTRIBUTION AND DEFORMATIONS IN THE HIP JOINT

Given model was used to perform a series of preliminary studies including the stress distribution and deformations with the use of Finite Element Analysis method.

FEA method

Finite Element Analysis (FEA) is one of basic methods of conducting a computer aided engineering calculations. It is one of the techniques of digitization of geometric systems, ie. dividing a continuum into a finite amount of subareas.

Main principle of FEA is to divide geometric model into finite elements uniting in nodes, which effects in creating discrete geometric model, split in simply shaped subareas, called the finite elements. During performing of calculations with the use of FEA other physical quantities are also being digitized: loads, tensions, restraints or other examples represented in the system with the use of continuous function. While performing the process of digitization software aims at maximally approximation of discreet and continuous form using approximation methods. After converting of the data analysis follows, consisting in uniting individual elements as a whole using equilibrium conditions and displacement compatibilities, which results in receiving a set of algebraic, simultaneous equations, posing as mathematical description of analyzed problem. Afterwards mentioned equations are being solved using values of equilibrium conditions, and their outcome used to compute sought quantities, ie. tensions [9, 8, 14].

Studies were performer in the environment of Solidworks Simulation software. Solidworks Simulation provides core simulation tools to test designs and make the decisions to improve their quality. The full integration creates a short learning curve and eliminates the redundant tasks required with traditional analysis tools. Component materials, connections, and relationships defined during design development are fully understood for simulation. Its main advantage is its cost, much lower than any other FEA software [9].

In order to achieve credible results, materials were assigned to each element. Properties of materials are presented in Table 1.

In order to receive correct results it was crucial to properly fix the model. For maintaining the best coherence with anatomical structure, the pelvic bone was fixed in the pubic symphysis and the sacroiliac (Fig. 3).

Next, the force was applied to the body of the femur with value matching the body weight (in N) of a person weighing approximately 80 kg, i.e. 800 N.

Table 1. Table of materials [2, 4, 7, 11, 13]

Material	Element	Young's modulus [MPa]	Poisson's ratio	Density [kg/m ³]	
Cortical bone	Pelvic bone	17400	0.39	1020	
Cortical bone	Femur bone	17600	0.3	1020	
Cartilage	Hip joint articular surfaces	Age 16-39	122	0.35	500
		Age 40-59	123	0.35	500
		Age 60-83	76	0.35	500

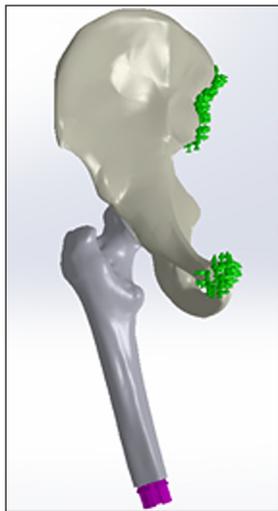


Fig. 3. Places of fixing of the pelvic bone (green) and application of force (violet)

Performed analysis shows, that articular cartilage indeed absorbs loads, which is proved by the fact, that most of the deformations happens right in this structure, with its peak value above the top of the head of the femur, directly where the spot of the highest value of load is expected. The scale of deformations, described with the von Mises equivalent strain (ESTRN), is presumably dependent on the value of Young's modulus of ar-

ticular cartilage, as seen on Figure 4 and Figure 5 or in Table 2 and Table 3.

According to Young's modulus of cartilage data negatively affects its ability to absorb loads generated from applied forces. Furthermore, in extreme cases of total destruction of articular cartilage caused by chondromalacia, loads are transferred directly from femur to pelvis, which causes excessive, even pathological deformations of both femur and pelvis, which are shown on Figure 4 and 5 (bottom right).

Table 2. Results of the analysis for femur

Young's modulus for femur cartilage [MPa]	The equivalent strain (ESTRN) [-]
76	5,135 e-02
122	3,182 e-02
123	3,156 e-02

Table 3. Results of the analysis for femur

Young's modulus for pelvis cartilage [MPa]	The equivalent strain (ESTRN) [-]
76	5,921 e-02
122	3,681 e-02
123	3,651 e-02

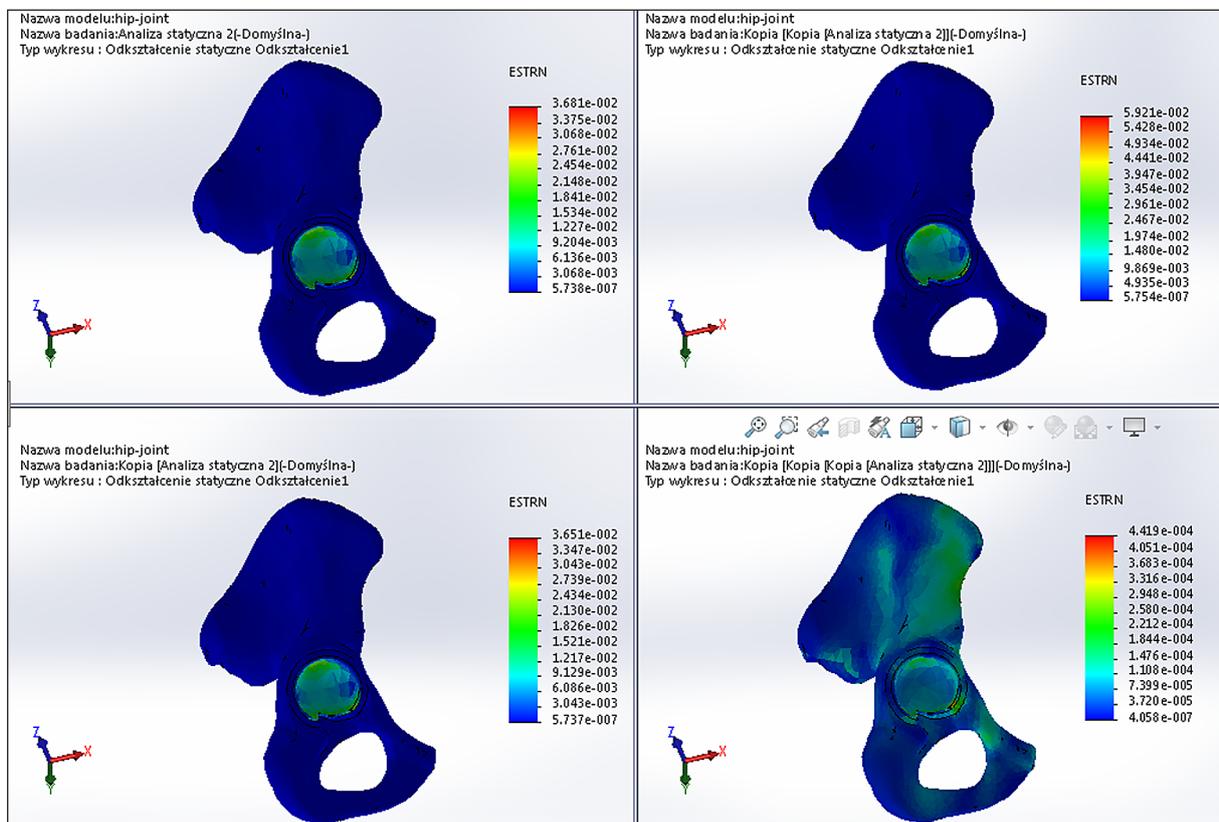


Fig. 4. Deformation of pelvis depending on properties of articular cartilage

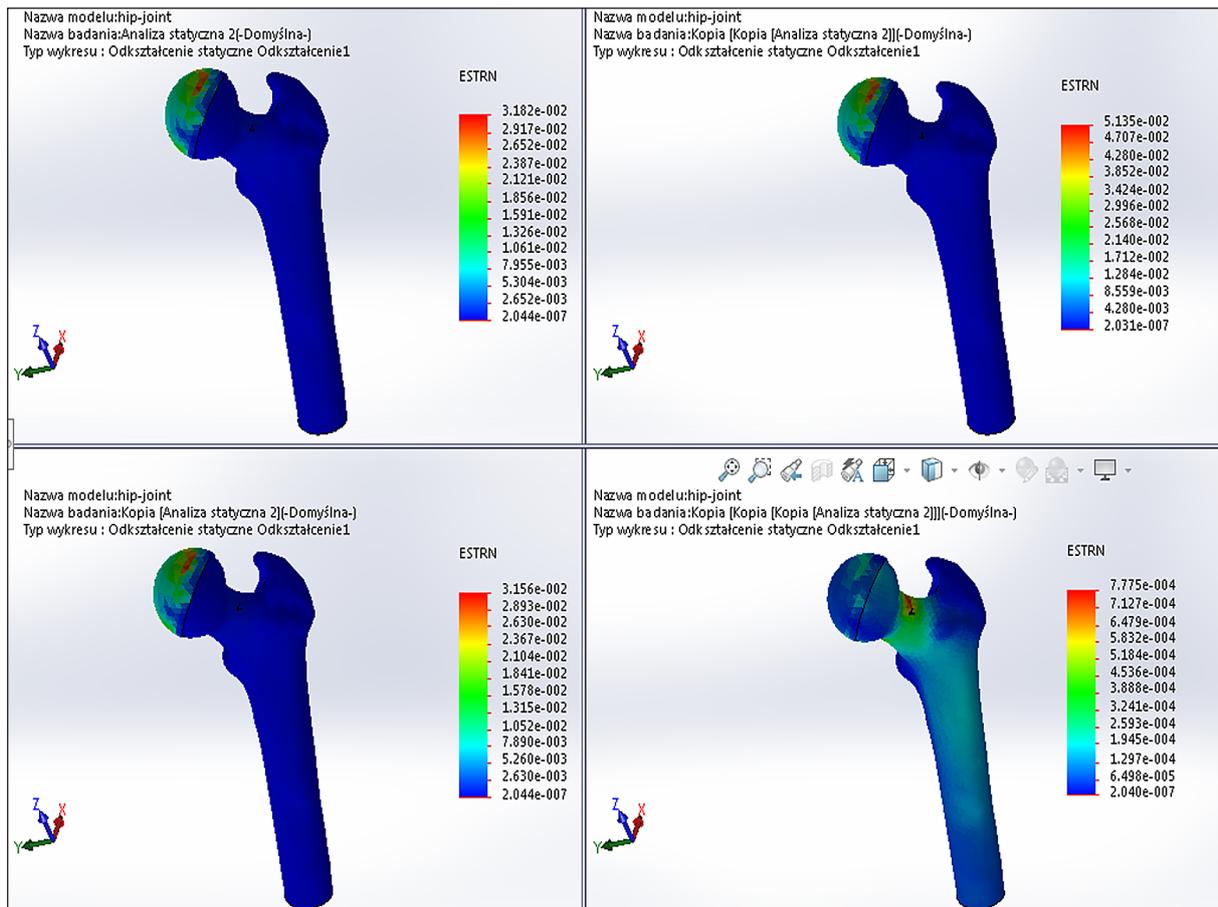


Fig. 5. Deformation of femur depending on properties of articular cartilage

CONCLUSIONS

The capability of absorbing loads by articular cartilage of the hip joint is preliminary determined as decreasing with increasing degenerations of the cartilage and with age of a patient. Without further information on changes of cartilage's mechanical parameters in time it is hard to determine the nature of relation between mentioned capability and these parameters. Given the way of obtaining results, method presented in the paper may provide additional informations about a condition of the hip joint, especially whether the progress of chondromalacia, without performing surgical procedures, which can be crucial for elder patients.

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