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A MULTI-SCALE STUDY OF THE HIP JOINT MECHANICS USING RIGID-BODY INVERSE DYNAMICS AND FINITE ELEMENT ANALYSIS

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ABSTRACT

Coxarthrosis (CA) is a non-inflammatory degenerative disease of the hip joint that provokes the destruction of the articular cartilage and bone growth, all in all resulting in pain and patient disability. Primary CA commonly affects old people due to cartilage ageing, but appears at an early age in about 20% of the cases, possibly caused by joint deformities and undue mechanical loads. However, deformities only explain up to 30% of the cases of juvenile CA [1].

Recently, it was suggested that juvenile CA could be caused by pathological loads transmitted to the hip joint cartilage, even in patients with apparently normal hip morphologies. In particular, Sánchez et al. [2] developed different finite element (FE) models of the hip joint, and their results strongly suggested that specific combinations of normal range anatomical angles provoke high cartilage stresses during daily activities. However, the confirmation of these results depends significantly on the boundary loads applied onto the FE model.

Hence, the goal of this study is to calculate the pressure within the cartilage during a patient's activity (in this case gait), based upon the movement kinematics and foot-ground contact forces captured in the laboratory. A multi-scale approach was developed for this purpose, which combined a musculoskeletal rigid body model of the entire human body and a FE model of the hip joint with a deformable cartilage. The process required acquiring proper boundary conditions for the FE model from the data calculated with the rigid body model. The workflow of the whole process is shown in Fig. 1.

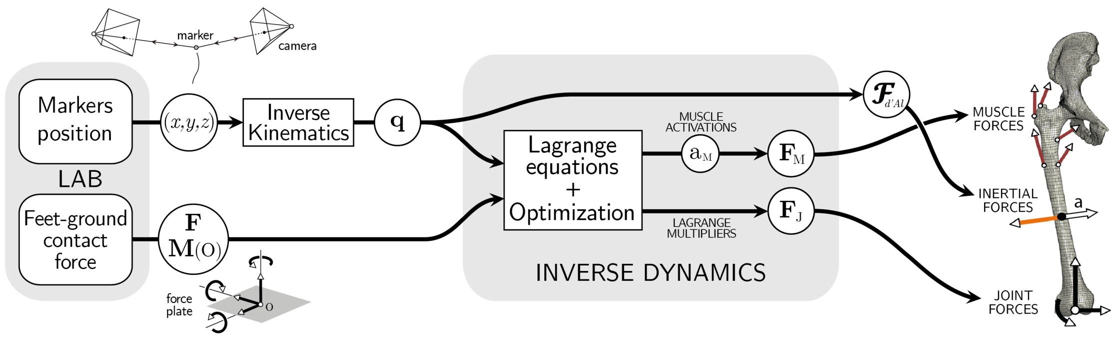

Figure 1. Workflow of the multi-scale modelling of the hip joint mechanics.

Both the rigid body and the FE models corresponded to healthy subjects. In order to capture the subject's movement, a set of 21 markers were placed on a volunteer and their trajectories were measured by means of 14 infrared cameras. The foot-ground contact forces were recorded using 2 force plates. All data was sampled at 100 Hz. The captured movement was reproduced on the rigid body model, which consisted on 12 segments (HAT –head, arms, trunk–, pelvis, 2 thighs, 2 shanks, 2 calcaneus, 2 talus and 2 toes) and had 23 degrees of freedom. The model was implemented on the OpenSim software [3]. The muscle forces F_M were estimated solving an inverse dynamics problem together with an optimization approach that minimized the sum of squares of muscle activations. Finally, Lagrange multipliers allowed to obtain the joint forces F_J.

For the FE calculations, a previously developed osteoligamentous hip joint model was used [2]. Bones were linear elastic, and cartilage tissues were considered as hyperelastic Neo-Hookean materials, with material parameter values taken from [4].

The boundary conditions for the FE model were calculated with the coordinates $\mathbf{q}$, the forces of the muscles attached to the femur (F_M) and the knee joint force (F_J) and torque (M_J). To avoid singularity in the FE system to solve, the pelvis was immobilized and a distribution of inertial forces was applied to the femur. These inertial forces emulate the movement of the bone at a specific instant, see Fig. 1. The d'Alembert inertial force $\boldsymbol{\mathcal{F}}_{d'Al} = -m\mathbf{a}$ was implemented as a body force applied to the femur elements, whose value depends on their location. Therefore, the body force $\mathbf{B}$ considering the gravity force was given by:

$$\mathbf{B} = \rho(\mathbf{g} - \mathbf{a}) \quad (1)$$

where ρ is the bone density, $\mathbf{g}$ is the gravity acceleration vector and $\mathbf{a}$ is the element acceleration.

The FE model was constituted by the articular cartilage, the femur and pelvis bones with the average values for a healthy subject of the anatomical angles: cervical-diaphyseal 120º, femoral anteversion 10º, acetabular anteversion 15º. The mechanical model for the cartilage was an incompressible hyperelastic Neo-Hookean solid with a shear modulus G = 6.8 MPa.

Finally, a static simulation of the FE model, with the described boundary loads, gave the pressure within the cartilage at the chosen instant of the movement captured at the laboratory. In order to verify the entire process, the pressure was calculated at the instant of the gait cycle when the hip joint force was higher; resulting in a pressure distribution physiologically consistent with a peak value of 3.5 MPa, see Fig. 2. This value is within the range of values that allow homeostatic response of chondrocytes under repeated mechanical loading. Moreover, boundary loads could be calculated over time in order to obtain the pressure within the cartilage during a time interval. In addition, these models could be modified in order to match with the subject morphology.

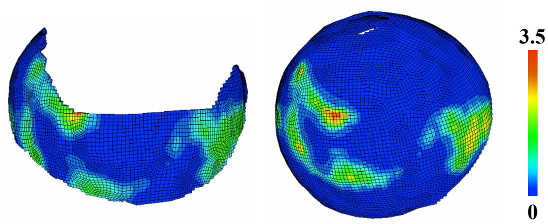

Figure 2. Distribution of the hydrostatic pressure, in MPa, within the hip joint cartilage (top view).

Hence, this analysis with two different models represents a first approach to a clinical diagnosis method and it will be a guideline for future studies, such as the determination of the juvenile CA. The presented method will allow the cartilage pressure to be estimated according to the patient morphology and his movement.

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