Multiscale analysis of the hip joint: translate mechanical information from inverse analyses of body motion into boundary loads for finite element calculations of organ/tissue biomechanics

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Hip joint deformities and undue mechanical loads may cause Juvenile Coxarthrosis, but explain only up to 30% of the cases. Recent finite element (FE) analyses suggested that increased load magnitudes could be transmitted to the cartilage in apparently normal joints but with specific combinations of anatomical angles [1]. However, the confirmation of these results depends on the boundary loads applied onto the FE model. Hence, this study aimed to calculate the pressure within the cartilage based upon patient movement kinematics.

A multi-scale approach combined a musculoskeletal rigid body (RB) model of the entire human body, with a FE model of the hip joint with a deformable cartilage. Both models represented a healthy subject. The RB model consisted in 12 segments (Head-Arms-Trunk, pelvis, 2 thighs, 2 shanks, 2 calcaneus, 2 talus and 2 toes) with al total of 23 degrees of freedom. It was animated through the trajectories of 21 markers, placed on a volunteer and captured by 14 infrared cameras. Foot-ground contact forces were recorded by using 2 force plates. Muscle forces F_M were estimated by solving an inverse dynamics problem together with a minimization of the muscle activations. Eventually, Lagrange multipliers allowed obtaining the joint forces F_J (Fig.1 Left).

For the FE calculations, a previously developed osteoligamentous hip joint model was used [1]. Bones were linear elastic, and cartilage tissues were considered as hyperelastic Neo-Hookean materials [2]. The boundary conditions were calculated according to the coordinates (q), the forces of the muscles attached to the femur, and the forces and torques (M_J) of the knee joint, extracted from the RB calculations (Fig.1 Left). The pelvis was fixed, and a distribution of d'Alembert inertial forces $\boldsymbol{F_{d'Al} = -ma}$ was applied to the femur elements as body forces, *B*, in order to simulate the movement of the bone, characterized by the acceleration vector, *a*, at a specific instant:

$\boldsymbol{B = \rho(g - a)}$ **,** where ρ is the bone density, and g is the gravity vector.

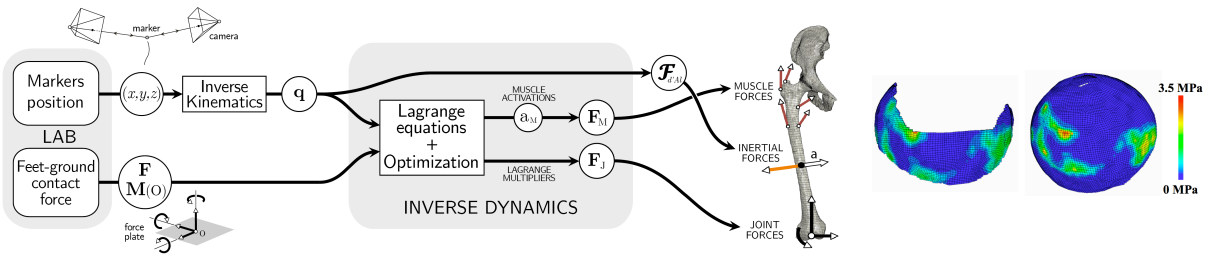

Fig 1: Left – Workflow of the multi-scale modelling of the hip joint mechanics. OpenSim and Abaqus simulations were combined through user-coded Matlab routines. Right – Hydrostatic pressure distribution in the articular cartilage layers of the FE model (cranial view)

Cartilage pressure within the FE model was calculated at the heel strike instant, resulting in a load distribution physiologically consistent with a peak stress of 3.5 MPa (Fig.1 Right). Calculated pressures were within the range of values that allow homeostatic response of chondrocytes under repeated mechanical loads [3].

This analysis with two different models represents a first approach to new clinical assessments of cartilage pressure, according to both the morphology and the mobility of the patients.

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