A LOWER LIMB MODEL VALIDATED THROUGH IN VIVO MEASURED JOINT CONTACT FORCES

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INTRODUCTION
Musculoskeletal models can estimate the internal forces acting in biomechanical systems with results allowing investigation of the relationships between bones and muscles. The muscle forces can be estimated by applying optimization methods, while the joint contact forces are computed by summing the muscle contributions to the intersegmental forces. On a lower limb perspective, the hip contact forces (HCFs) originate from the contact between the acetabulum and the femoral head and knee joint contact forces (KCFs) can also be defined for the tibio-femoral joint. The purpose of this work is to present a lower limb model entirely implemented in OpenSim [1] and make an attempt to validate this model with respect to the experimental joint contact forces measured previously by means of instrumented prostheses at the hip [2] and at the tibio-femoral joint [3-4].

METHODS
A lower limb model based on a recently published anatomical dataset [5] was built in OpenSim [1] in order to investigate some of the most common daily living activities. The unilateral model includes 6 segments treated as rigid bodies (pelvis, thigh, shank, patella, hindfoot and midfoot-phalanxes) and it is visible in Figure 1. Pelvis and thigh are connected by a spherical joint (hip), thigh and shank are linked by a hinge joint (knee) and the ankle joint complex, composed of the talocrural and the subtalar joint, is modeled by two hinges. When knee flexion occurs, the patella is dragged by the patellofemoral ligament (assumed inextensible) and rotates around an axis embedded in the distal part of the femur. The model includes 163 actuators representative of 38 muscles. Where appropriate muscle paths are enhanced by wrapping surfaces and via points. Muscle dynamics has not been included in the model at this stage of development, although it has been shown not to influence muscle force estimation when walking is investigated [6].

The experimental kinetics and kinematics of a patient that underwent total hip replacement (THR) contained in the publicly available dataset HIP98 [2] were used to simulate walking (8 trials) and stairs climbing (6 trials). The patient is named HSR in the dataset and the choice of the investigated activities is justified by walking being the most frequent daily activity and ascending stairs one of the most critical in terms of both HCFs and KCFs. After scaling the general model to the patient anthropometric dimensions, the static optimization technique was applied in order to estimate muscle forces. This approach consists in minimizing a physiologically meaningful function of the muscle forces under the constraints of moment equilibrium at the joints. In this work the sum of the cubed muscle forces normalized with respect to the maximal isometric force

\[ \sum \left( \frac{F_i}{F_{iso,i}} \right)^3 \]

was minimized for \( i = 1,2,3,...,n \), where \( n \) is the number of muscle actuators included in the model. This function has been previously related to maximal muscle endurance [7].

Following prediction of the muscular loads, the joint contact forces could be determined. The calculated HCFs and KCFs were used to attempt a validation of the model, as both these internal forces have been previously measured in patients by means of instrumented prostheses [2-4].

A limitation of the proposed methodology is the use of THR patient kinematics and kinetics to evaluate KCFs measured in total knee replacement patients. This choice is justified by the current lack of availability of TKR patients kinematics in the literature and by the possibility of having at least a cycle-to-cycle comparison for the HCFs. A comparison in terms of peak values and timing of the numerical HCFs against the synchronous experimental measurements available in the HIP98 dataset [2] is reported, while just a qualitative assessment of the calculated KCFs against the values recorded by instrumented prostheses in other patients [3-4] is possible.

RESULTS AND DISCUSSION
The HCFs for the investigated patient are visible in Figure 2 for normal walking (average speed = 1.15 m/s) and stair climbing. The average relative deviation per cycle of the numerical and experimental peaks with respect to the experimental value of the force is 34.7% BW for walking and
16.1% BW for stair climbing. The timing of the HCF peaks is well correlated with the experimental values (average peak time shift is 2.7% of the total gait cycle and 3.7% of the total ascending cycle).

The KCFs are visible in Figure 3 as individual trials for the same activities of the same patient. A comparison with the available in vivo measurements is possible in terms of magnitude of the total force and timing of the force peak. The peak range of the total force acting on the tibia is between 303-380% BW for walking (average value 338% BW) and 264-313% BW for stair climbing (average value 282% BW), while the ranges measured by Kutzner et al. in five patients is between 220-295% BW for walking (average value 261% BW) and 275-360% BW for stair climbing (average value 316% BW)\[3\]. The values recorded by D'Lima et al. \[4\] for a single patient (230% BW for walking and 300% BW for stair climbing) are in the range reported by Kutzner et al. \[3\].

In our simulations the KCF peaks occurred close to the contralateral heel strike and to the foot contact with the stair step, which is consistent with measurements in the literature \[3\].

CONCLUSIONS
A three dimensional lower limb model has been developed in OpenSim and a validation using hip contact forces for the two most frequent activities of daily life has been attempted for a single patient. A promising agreement of the estimated HCFs has been calculated in terms of peak values and timing for the two activities investigated with respect to the experimental measured values. On the other hand, for the same tasks, the forces calculated at the knee suggested an inadequacy of a simple hinge model in representing the knee joint, as KCF magnitude is not consistent with the existing literature reporting in vivo measured values. Nevertheless, the conclusions obtained in this case study need to be confirmed by the results of additional simulations based on a more extensive set of patients and activities.

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REFERENCES