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FALSIFICATION OF A LOWER LIMB MODEL PREDICTING HIP CONTACT FORCE VECTORS

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INTRODUCTION

Musculoskeletal models are numerical tools of great potential to estimate internal loads of the human body, but they still suffer from a lack of performance assessment that is preventing them from being considered as reliable as any other CAD software [1]. Attempts at validation of the hip contact force (HCF) predicted by lower limb models based on comparison against measurements from instrumented prostheses are available in the literature [2, 3], although it has been reported that the direction of the HCF in the transversal plane tends to be predicted less accurately than in the sagittal plane [2, 4].

As the directionality of the HCFs is paramount in applications such as edge loading, the purpose of this investigation is to assess if a musculoskeletal model of the lower limb can actually produce the HCFs that have been measured experimentally for the same kinematics and kinetics [5]. Challenging the model in these terms can be considered a falsification strategy [6].

METHODS

A general musculoskeletal model of the lower limb based on the Klein Horsman dataset [7] was implemented in OpenSim [8] and scaled in order to represent the four patients included in the HIP98 dataset [5]. The unilateral model includes six body segments (pelvis, femur, tibia, patella, hind-foot and mid-foot plus phalanxes) connected by five joints (pelvis-ground joint: six dofs, hip: spherical joint with three rotational dofs, knee, talocrural and subtalar joints: rotational joints with one dof each) and 163 actuators representing 37 muscles, whose paths are enhanced by means of via points and wrapping surfaces. The maximal force of each actuator is proportional to the muscle cross sectional area (PCSA) shared between the number of actuators representing the muscle multiplied by a constant termed muscle tetanic stress.

For the purposes of this investigation, the model was modified as follows (Figure 1, A):

1) the pelvis segment was removed and the femoral kinematics described by six dofs with respect to the ground.
2) pelvic muscle attachments and related via points and wrapping surfaces were defined using splines in order to reproduce the assigned pelvic kinematics with respect to the ground.

As the patella moves as a function of the knee flexion angle [7], the model kinematics could be described by eight generalized coordinates.

The HCFs measured by Bergmann et al. [5] were applied directly to the femoral head (Figure 1, A), so that the hip crossing muscles were required to provide exactly the total amount of force necessary to equilibrate the joint contact force (Figure 1, B). In other terms, they were constrained to equal a specific vector equilibrating the imposed HCFs (applied as an external load); the three constituent scalar constraint equations were named directional constraints.

The existence (or non-existence) of at least one combination of muscle forces capable of reproducing the measured hip contact forces was checked for each frame of two simulated activities (level walking and stair climbing) within the space of the solutions of the equations of motion by solving a linear programming problem having constant objective function and constrained by the equations of motion. The effect of including individual directional constraints and their combinations was assessed with respect to the total percentage of solvable frames.

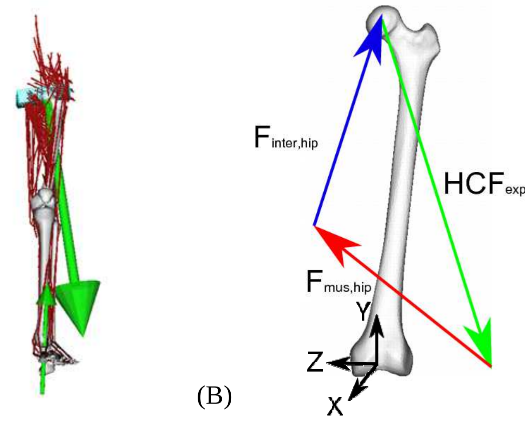

Figure 1: (A) Representative image of the musculoskeletal model used in this investigation. (B) Representation of the femoral coordinate system and muscle and intersegmental force contributions to the hip contact force vector.

The sensitivity of the results to the tetanic stress was investigated by varying its value in steps of 10 N/cm^2 between 30 N/cm^2 and 150 N/cm^2, while the effect of possible errors in the estimation of the intersegmental moments (or muscle moment arms) was assessed by modifying their nominal value (calculated from inverse dynamics analysis performed in OpenSim) in 10% steps to ±30%.

RESULTS AND DISCUSSION

When the directional constraints were not imposed 100% of the analyzed frames were solvable for all subjects and activities in the considered muscle tetanic stress range (Figure 2). If directional constraints were also considered, the number of solvable frames was minimum when constraints for all directions were included ('X+Y+Z' in Figure 2) and increased with larger values of tetanic stress; the range of percentage of solvable frames was 43-84% for level walking and 28-88% for stair climbing, across the four considered subjects.

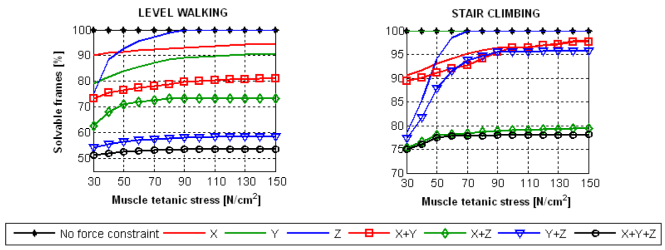

Figure 2: Percentage of kinematics frames (for a representative subject) for which the equilibrium equations present feasible solutions and their sensitivity to the muscle tetanic stress when the net contribution of hip muscles is not specified ('No directional constraint'), is specified in a single direction ('X', 'Y' and 'Z') and is specified in multiple directions ('X+Y', 'X+Z','Y+Z','X+Y+Z').

As expected, the percentage of solvable frames also increased when decreasing the intersegmental moments (Figure 3), reaching up to 98% for level walking and 100% for stair climbing for one specific subject (all directional constraints included).

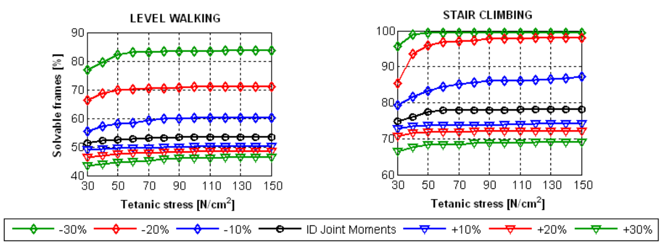

Figure 3: Percentage of kinematics frames (for a representative subject) for which the equilibrium equations present feasible solutions and their sensitivity to the muscle tetanic stress when the intersegmental joint moments calculated through an inverse dynamics analysis are varied in steps of 10% to ±30% of the original value.

The space of the solutions of the equilibrium equations is a vectorial space and can be explored in different ways, for instance to evaluate suboptimal neuromotor control [9], but in the subset defined by meaningful values for the muscle forces ($0<F_i<F_{i,max}$) it does not necessarily contain a muscle force vector capable of yielding the measured HCFs while providing the joint moments necessary to satisfy the equilibrium equations at the rotational dofs. The frames for which a solution could not be found generally corresponded to the stance phase of gait.

Including stronger muscles in the model, i.e. broadening the domain of the space of the solutions, increased the percentage of solvable frames until a plateau value (lower than 100%) was reached; while variations of the intersegmental joint moments changed the value of this plateau so that a larger percentage of frames were solvable (even all of them for one of the subject) when lower joint moments were considered.

Imposing the constraint on the medio-lateral HCF component ('Z' in the Figure 2) in combination with any of the other constraints dramatically decreased the number of solvable frames, suggesting that the geometrical representation of the gluteal muscles (especially gluteus medius, active during the stance phase of gait) needs to be improved in the model. Also, the subject having the minimum number of solvable frames presented the larger femoral anteversion of the four, so suggesting that including subject specific bone geometries and muscle attachments in the model is a necessary condition to predict realistic HCF vectors, especially when there are prominent variations from the general model.

CONCLUSIONS

The falsification strategy presented above leads to an appreciation that some limitations in predicting HCFs can be intrinsic to the adopted musculoskeletal model geometry as opposed to related to the methodology used to resolve the muscle load distribution problem, e.g. static optimization, computed muscle control or EMG-based techniques. The main limitation of this technique is that it requires a dataset including joint contact forces and synchronous kinematics and kinetics, which are generally not available except for few valuable exceptions [5, 10].

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