Hip compression force estimation with a comprehensive musculo-skeletal model

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Introduction

Muscle forces exerted during human movement provide insight in tissue load and muscle function and/or malfunction. The use of musculoskeletal models in the estimation of muscle force has been widely reported in the literature (see for review [1]). Although model-based estimation of muscle force showed to have clinical potential, many challenges still have to be overcome.

(1) In many models muscle force can instantaneously drop to zero or rise to maximal force (e.g. [2]) However, in reality muscular dynamics will prevent such fast transitions in force. (2) Objective functions based on mechanical measures such as muscle force or stress are frequently used (e.g. [3]). However, the validity of these functions is unknown. A recent study posed an energy-related cost function that had a better correspondence with muscle energy consumption than cost functions based on muscle stress [4]. (3) Since the objective function and its boundaries are a function of anatomical parameters, the outcome of a muscle force optimization is highly dependent on such data. Frequently used datasets are either incomplete or constructed by merging of datasets based on different individuals. This might result in possible inaccuracies caused by inter-individual anatomical differences and co-variance between parameters.

We have developed a musculoskeletal model of the lower extremity based on a recently collected, extensive and consistent anatomical dataset [5]. In the model muscle forces are optimized using a recently developed inverse-forward dynamic optimization (IFDO) method [6], which takes muscle dynamics into account. A recently proposed energy related cost function is used that showed to have a good correspondence with energy consumption [4].

The aim of this study is to evaluate the effect of IFDO and the cost function on hip reaction forces determined with the model, by comparing with existing methods.

Methods

In this two-legged model, 10 joints are crossed with 264 Hill-type muscle elements, defined by muscle parameters such as optimal fiber length. 'Via' points or wrapping geometries were defined in case of a curvature in muscle line of action [5]. Muscle dynamics are described by a third-order muscle model with excitation, active state and contractile element length as state variables. To evaluate the dynamic model properties, gait kinetics was collected for a healthy male subject (age 21, mass 85 kg, length 1.85m). The size of the model and the muscle parameters (e.g. optimal fiber length) were scaled to the length of the segments of the subject. PCSA was scaled to subject mass. The objective function represents the muscle energy consumption of the two-major energy processes in the muscle. The first is the detachment of cross bridges which depends on fiber length and the muscle force. The second process is the re-uptake of calcium which depends on muscle mass and the ratio of actual muscle force and maximal force at a certain length. IFDO was used to estimate muscle forces during gait. This method, to solve the load sharing problem, accounts for muscular dynamics. For each time step at a given state, the minimal and maximal possible muscle force was determined, resulting in respectively a lower and upper limit. These boundaries were used as a constraint, preventing instantaneous drop to zero or increase to maximum in optimized muscle force. After the muscle forces are estimated, an inverse model is used to update the states of the muscle model. In order to satisfy the equations of motion, a second constraint enforces the contribution of the muscles to the joint moment to equilibrate the calculated joint moment. The joint forces are estimated as the combined result from all muscles crossing the joint.

Results and discussion

In agreement with previous model simulations [7], variation in mechanical based cost functions had a small effect on hip compression force. However, in addition, our simulations showed that when the energy related cost function was used instead of a mechanical based function, hip compression force increased with 30%. An optimization with dynamic muscle force constraints resulted in 70% increase in compression force in the hip when compared to an optimization with static constraints. This is in contrast with a previous study showing that static and dynamic optimization solutions are practically equivalent [8].

The current model is based on accurate and consistent anatomical data, which will likely improve the outcome of the model. In this study is showed that besides this effect, IFDO and the energy cost function have a substantial influence on predicted hip forces. It is expected that this will improve the accuracy of joint load estimation in comparison to commonly used methods that exclude muscle dynamic properties as described earlier. However, since the actual hip compression force of the subject in this study was not measured directly, this remains speculative.

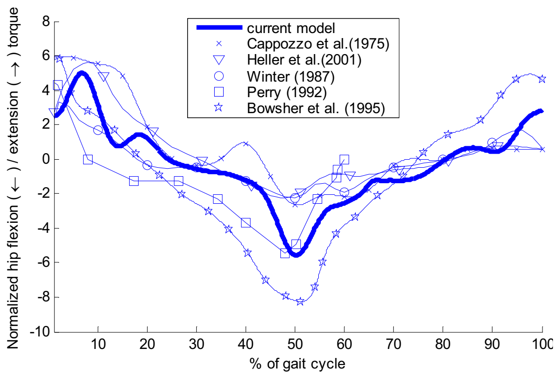

Figure 1. *Hip flexion moment determined with the model normalized to subject length and mass as a function of gait cycle in comparison with normalized moments from the literature*

In the literature a wide range of hip compression forces is reported as a result of differences between subjects e.g. walking speed, style) and used methodology (e.g. # optimized DOF, optimization method, cost function). In general, calculated joint moments, which directly effect hip

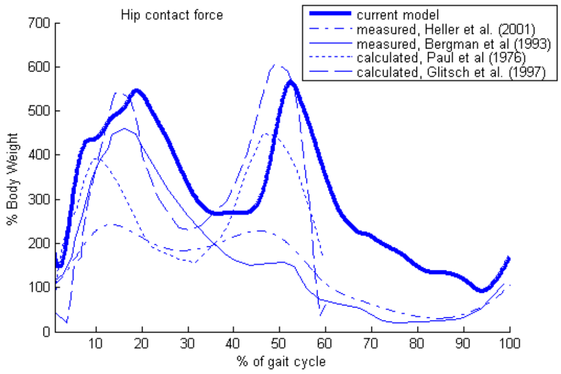

Figure 2. *Hip compression force (% BW) determined with the model as function of gait cycle in comparison with model simulations and in vivo measured compression forces from the literature.*

compression force, fall reasonably well within the wide range found in the literature (See as example hip flexion torque in figure). Despite normalization to subject length and weight, a large variation in amplitude was found between different studies. Such variations can be mainly attributed to differences in walking speed and style for example due to age. The hip compression force determined with the model using IFDO and the energy related cost function was around 2 times larger as measured in an in vivo study [9] (figure 2). This difference is in agreement with the large difference in hip moments between these studies (1.9 vs. 5.5 normalized peak hip flexion moment as shown in figure 1) as a result of lower walking speed (1.08 instead of 1.51 m/s) and subject condition (61 year old with a hip prosthesis instead of healthy 21 year old subject).

Conclusions

Besides the effect of anatomical data on model output [5,6], this study shows that the estimation of hip compression force is highly dependent on the used optimization method and cost function. This emphasizes the relevance of the use of accurate cost functions and optimization methods in order to estimate accurate muscle forces and joint loads.

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