Towards the development of a simulation framework to assess and enhance crutch-assisted gait

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ABSTRACT

1 Introduction

Crutches are widely used to assist gait in individuals with lower limb impairment. Walking with crutches alters both upper and lower body loading, potentially leading to discomfort and injury [1]. As such, it is important to study how crutch-walking affects upper and lower extremity movement patterns under different conditions, e.g., altered crutch lengths or gait pattern. Improper crutch length fitting may result in upper limb pain and injury. Computer modelling and simulation can help predict how varying parameters in a gait aid may affect patient outcomes, without the need to expose patients to tiring gait lab experiments.

We recently developed an optimal control prediction framework to predict crutch-assisted gait patterns with a three-dimensional ideal-torque driven model [2]. The framework successfully predicted different gait patterns (four-point, two-point and swing-through), and the effect of changing axillary crutch length on upper limb kinematics during swing through gait generally matched experimental observations [3]. Yet joint kinematics, especially at the ankle, were not well predicted, mainly because ideal torques were not able to represent physiological joint torques. In order to include muscle forces in the model, without adding excessive complexity, it has been proposed to use physiological muscle torque actuators (MTGs) to represent the resultant torques being generated by muscle forces [4].

The main goal of this study was to enhance our crutch-assisted walking prediction framework to improve the realism of the predicted gait patterns. Specifically, our aims were: (1) to predict three-point crutch walking, (2) to add physiological muscle torque generators to actuate the model, and (3) to evaluate if the model was capable of predicting realistic crutch walking patterns under different conditions (e.g., different speeds and stride lengths).

2 Methods

A musculoskeletal model comprising 21 degrees of freedom was employed for the simulations. The foot-ground contact model (CM) consisted of a viscoelastic contact model for the normal force and a simple continuous friction model addressing tangential forces. In contrast, the crutch-ground CM was simplified to a viscoelastic contact model along the crutch's axis. The hand-crutch contact was modelled as a welded joint, directly transferring forces from the crutch-ground CM. The gait of this model was predicted on GPOPS-II, using the direct collocation method in a single-phase optimal control problem and an implicit formulation of the musculoskeletal dynamics. The cost function consisted of a combination of mechanical power, residual forces and muscle activation and joint jerk (derivative of acceleration) minimization. Regarding the boundary conditions on the state variables, they were taken from experimental unassisted gait measurements of different subjects.

To implement the new gait pattern, the maximum force output of one foot was limited, and the crutches were virtually coupled with it, forcing the model to distribute the weight across all three contact points. Then, muscle torque generator functions, including angle-dependent, angular velocity dependent and passive torque production at each joint, were calibrated using experimental data from a young healthy male collected with a BIODEX dynamometer equipment [5]. Finally, to study the effects of spatiotemporal parameters on the crutch-assisted gait pattern, we performed several simulations, for a variety of mean walking speeds (between 0.6 and 1 m/s), and for a variety of stride lengths (between 0.8 and 1.2).

3 Results and discussion

The desired improvements were successfully implemented to the simulation framework, at a cost of nearly doubling the simulation time. When comparing the MTG and torque-driven approach, the ground reaction forces predicted by the model appeared physiological, with the model distributing the weight similarly between the two crutches and the injured feet. Regarding shoulder and lumbar joint torque prediction, the MTG-driven model performed similarly than the more ideal-torque-driven model (Fig. 1 left, top), with an average difference of 2.0 and 13.3 Nm and maximum difference of 19.3 and 32.6 Nm, respectively. Nevertheless, the MTG-driven model better predicted the initial shoulder torque, which is commonly hard to simulate, yielding a torque that was 22% smaller compared to the predicted by the ideal-torque-driven model. Regarding joint coordinates (Fig. 1 left, bottom), the model driven by MTGs was capable of predicting a more upright position, in general for all simulations. In summary, adding MTGs to the model resulted in a slightly improvement of the predicted gait pattern. More research is needed to compare more in depth both modelling approaches. In future work we will use MTGs to personalise muscle group weakness for specific subjects, which may not be possible using ideal torques.

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The simulation outputs showed that the framework was also capable to predict gait under different conditions, to get detailed insights about the joint loading. For this proof of concept study, where we varied speed and stride length (Fig. 1, right), we could observe that higher speed usually lead to higher torques in the lumbar load, and that for a fixed speed, it peaked out at around 1 m/s. Regarding the shoulder load, it could be seen that at higher speeds, an increase in stride length yielded a reduction in joint loading, as a higher stride translated to a reduction in cadence.

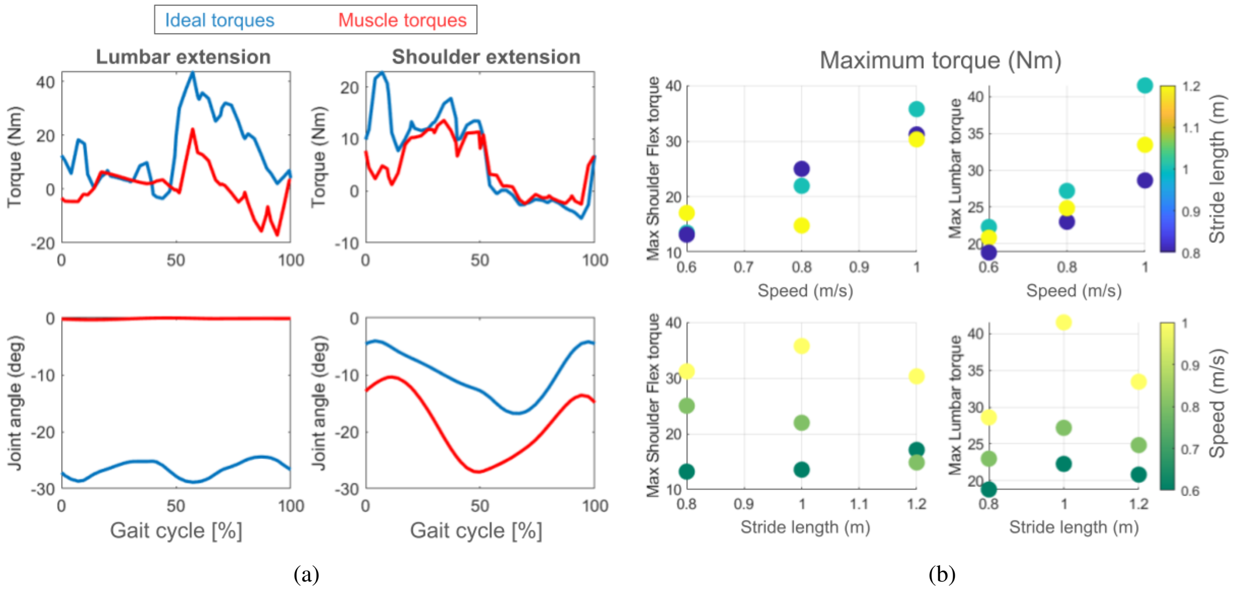

Figure 1: (a) Comparison of shoulder and lumbar torque and angle between ideal-torque-driven (blue) and MTG-driven (red) simulation. (b) Shoulder and lumbar torques under different speed and stride length combinations.

4 Conclusion

This study shows how this simulation framework can successfully accept new gait patterns and more complex muscle actuators, while keeping a high degree of realism. Furthermore, with the shown predictions we can observe how interconnected parameters like speed and stride length are and how much they can impact joint loading. On the long term, the availability of an algorithm that allows the prediction of crutch walking patterns could be useful to study the impact of different conditions on crutch walking (such as type of crutch, walking pattern and different spatiotemporal parameters). This algorithm, together with neuromusculoskeletal models personalised to specific patients with mobility impairments (such as spinal cord injury subjects), could be used in the future to define rehabilitation treatments aimed at maximising recovery of walking function.

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