Predictive simulations match the observed effects of changing axillary crutch length on upper limb kinematics during swing-through gait

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EXTENDED 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. Improper crutch length fitting may result in upper limb pain and injury. Kuntze et al. have recently studied the effects of axillary crutch length on upper limb kinematics during swing-through gait [2]. They found that crutches longer and shorter than the standard fit resulted in altered kinematics across all joints, especially in the shoulder.

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. For this reason, the availability of an algorithm that allows the prediction of crutch walking patterns could be useful in order to study the impact of changing conditions on crutch walking, and could overcome some limitations of experimental studies, such as difficulty in recruiting subjects or limitation in the number of tests that can be performed. The main goals of this work were to use predictive simulations to a) study the effects of changing crutch length on the kinematics of a healthy subject performing axillary crutch-assisted swing-through gait and b) to compare those effects with the ones observed experimentally [2].

2 Methods

Gait data were collected from a healthy young male (24 years old, height 1.72 m, weight 82.55 kg) who performed 10 swing-through axillary crutch-assisted gait trials landing on the left leg. Marker trajectories were collected for 63 markers, and ground reaction force data were recorded for the left foot and the left crutch. A 3D full-body torque-driven model of the subject using axillary crutches was created starting from a published OpenSim (National Center for Simulation in Rehabilitation Research) model [3]. The model possessed 35 degrees of freedom (DOF), and was scaled to the subject using a neutral trial and the OpenSim scaling tool. Each axillary crutch was introduced into the model as a rigid body welded to the corresponding hand segment. Nine different models were developed, where crutches were shortened and lengthened by 1, 2, 3 and 4 cm starting from the baseline crutch length (1.24 m). Foot-ground and crutch-ground compliant contact models were implemented in Matlab (MathWorks, Inc.) using visco-elastic force models, which were calibrated to match the collected data [4].

We performed predictive simulations (i.e., without tracking the experimental data) for each crutch length condition. The different optimal control problems were based on [4], and solved by a direct and simultaneous collocation method using the optimal control software GPOPS-II [5]. The cost function (Eq. 1) included two terms: minimization of angular momentum and minimization of the time derivative of joint torques:

$$J = \int_{t_0}^{t_f} \left(\sum_{i=1}^{n_b} ||\boldsymbol{L}_i||^2 + \sum_{i=1}^{n_q-6} \dot{\tau}_i^2 \right) dt \tag{1}$$

where t_0 and t_f are the initial and final simulation times, respectively; n_b is the number of rigid bodies in the model; n_q is the number of model coordinates or DOF; $\boldsymbol{L}_i$ is the local angular momentum at the centre of mass of the i^{th} body of the model; and $\dot{\tau}_i$ is the i^{th} component of the vector of joint torque change $\dot{\boldsymbol{\tau}}$.

The initial guess for simulations was defined as the mean pattern of all 10 trials, resulting from an inverse kinematics and inverse dynamics analysis using OpenSim. We assumed that crutch forces were symmetric (as force data were only available for the left crutch). We compared joint coordinate patterns obtained for the model with regular crutch length against those obtained for the models with shorter and longer crutch lengths. Changes in upper limb kinematics were qualitatively compared to the experimentally derived results [2].

3 Results and Discussion

The predicted gait pattern with standard crutch length followed the main features of the experimental swing-through gait pattern, but showed some differences in joint coordinates and ground reaction forces. The trends in lumbar extension and lumbar bending, as well as in shoulder flexion and knee flexion, were well-predicted. However, the range of motion (ROM) of shoulder flexion and ankle dorsiflexion was larger in the simulations, compared to the experimental data, and it was lower for knee flexion. Furthermore, the first peak in foot normal ground reaction force and the second peak in crutch normal ground reaction force were not predicted.

Regarding the effects of crutch length variation on upper body kinematics, we found that longer crutch length resulted in reduced shoulder flexion, shoulder adduction and wrist deviation, and increased elbow flexion (Figure 1). These results strongly agree with the results reported from the experimental study [2]. Shorter crutch lengths produced the opposite changes compared to larger lengths, but they had a smaller effect on upper limb kinematics.

We also studied the effects of crutch length variation on lower limb kinematics. Longer crutch lengths resulted in lower hip flexion and shorter crutch lengths resulted in higher hip flexion. There was no clear effect on hip adduction. Overall, the effects on knee flexion and ankle dorsiflexion were small for both longer and shorter crutches.

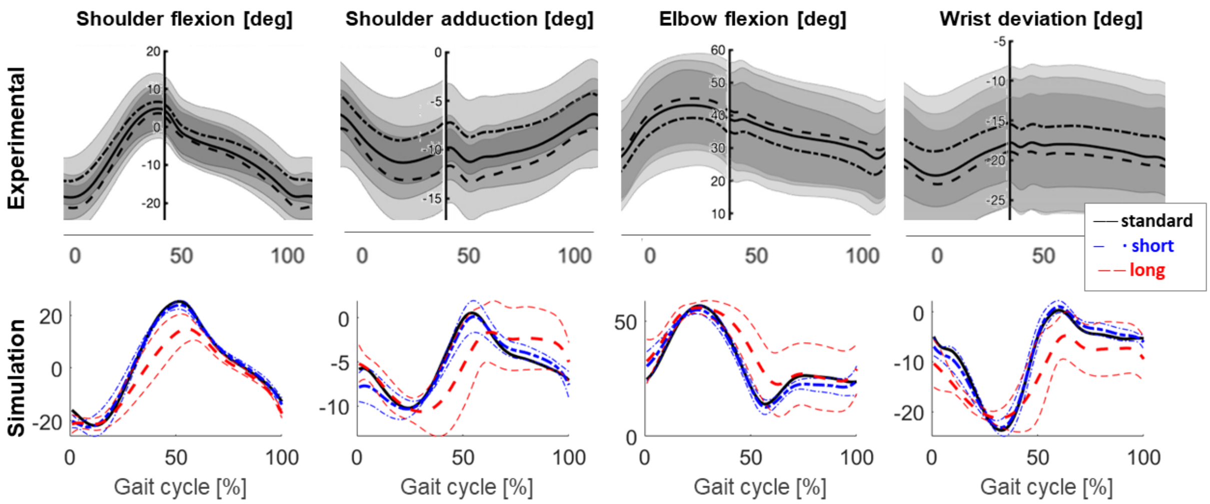

Figure 1: Mean and standard deviation of experimental and predicted upper limb kinematics. Results are shown for the left side, which was the side of the weight-bearing leg. [Top] Experimental results from 15 healthy subjects and three different crutch lengths (standard, 5 cm shorter and 5 cm longer). Adapted from [2]. [Bottom] Simulation results from 9 different simulations: baseline and crutches shortened and lengthened by 1, 2, 3 and 4 cm starting from the baseline crutch length.

4 Conclusion and Future Work

Our optimization formulation generally matched the experimentally observed effects of changing axillary crutch length on upper limb kinematics during swing-through gait. However, the predicted movement pattern for gait with the standard crutch length differed from the one captured experimentally. More research is needed to improve the foot and crutch contact model parameter values. As ankle kinematics are especially linked to contact forces, we hypothesize that having a contact model that represents the foot-ground contact more accurately will lead to better prediction of ankle kinematics. Moreover, we will extend the model to include muscle torque generators, to produce more physiological gait patterns.

Acknowledgments

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