A Reduced Gait Model for Motion Prediction in the Clinic

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Detailed musculoskeletal models have provided insight to the research community [1] but remain challenging to use in a clinical setting. Effective methods need to be developed to fit detailed musculoskeletal models to a specific subject before they can be employed in routine clinical work. However, not all clinical applications require a model with a one-to-one correspondence to the musculoskeletal system.

In this work we present an anthropomorphic model of a human that is actuated by a set of muscle-torque-generators (MTGs). The strength of an MTG can be fit to a specific subject using measurements of maximum-voluntary-contraction (MVC) at each joint. Although the MTGs cannot be used to estimate bone-on-bone contact forces, nor muscle fiber lengths, they can be used as part of an optimal-control-problem (OCP) to predict the motions, ground forces, and net joint torques of a subject's gait.

Our long-term goal is to improve the design and fitting of orthoses to children with pathological gait. Gait pathology is complex, and so we are motivated to first test this approach in a typically developing child. We compare our results to the experimental data of Schwartz et al. [2] for evaluation.

1 Methods

We model the human body as a 10 degree-of-freedom planar mechanism (Fig. 1). Foot-ground contact is modeled using kinematic constraints between the 2 points of each foot and the ground such that the normal forces are positive. The torque-angle $\mathbf{f}^{\mathrm{A}}(\theta)$ and torque-velocity $\mathbf{f}^{\mathrm{V}}(\omega)$ characteristics of the MTGs are expressed as $\mathscr{C}_2$ continuous 5^{th} order Bézier splines that have been fitted to the dynomometry data of Anderson et al. [3]. These characteristic curves are used to evaluate muscle torque

$$\tau^{\mathrm{M}} = \tau_{\mathrm{o}}^{\mathrm{M}} a \mathbf{f}^{\mathrm{A}}(\theta)\mathbf{f}^{\mathrm{V}}(\omega) \quad (1)$$

where a is the muscle activation. We use Thelen et al.'s [4] model of muscle activation

$$\dot{a} = \begin{cases} (e-a)\left(\dfrac{e}{\tau_{\mathrm{A}}} + \dfrac{1-e}{\tau_{\mathrm{D}}}\right) & \text{if } e \geq a \\ \dfrac{e-a}{\tau_{\mathrm{D}}} & \text{otherwise} \end{cases} \quad (2)$$

where e is electrical excitation from the nervous system. Winters and Stark's [5] data have been used to set the activation ($\tau_{\mathrm{A}} = 0.011$) and deactivation ($\tau_{\mathrm{D}} = 0.068$) time constants.

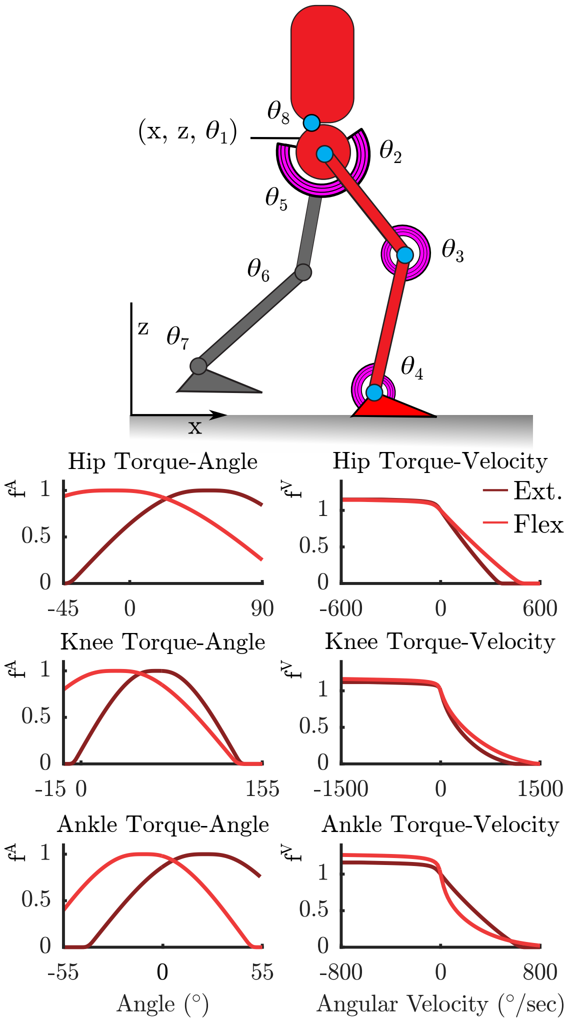

Figure 1: A 10 degree-of-freedom sagittal plane gait model and the characteristic curves that define the torque-angle (left column) and torque-velocity characteristics (right column) of the muscle-torque-generators.

The geometry and inertia of the model is fitted to a typically developing child (height 1.26 m, weight 25.9 kg) using Jensen's regression equations [6]. The strength of the model has been set using the maximum-voluntary-contraction data of Eek [7]. The values for $\tau_{\mathrm{o}}^{\mathrm{M}}$ for each joint

Joint	$\tau_{\mathrm{o}}^{\mathrm{M}}$ Ext. (Nm)	$\tau_{\mathrm{o}}^{\mathrm{M}}$ Flex. (Nm)
Hip	49	34
Knee	36	19
Ankle	39	13

have been found using the joint angle θ and τ^{M} from Eek et al.'s data, by setting $a = 1$, and solving Eqn. 1 for $\tau_{\mathrm{o}}^{\mathrm{M}}$. We have set the strength of the lumbar joint to be equal to the hip as we lack data for this joint.

To predict the walking pattern of the model we formulate a multi-phase optimal-control-problem (OCP) with the goal of minimizing

$$\min_{\underline{x}(\cdot),\underline{u}(\cdot),\underline{v}} \quad \frac{\sum_1^4 \int_{v_{j-1}}^{v_j} \underline{a}(t)^T \underline{a}(t)\,dt}{r(T)} \tag{3}$$

the integral of activation-squared-per-distance-traveled across all 4 of the problem phases. Dividing by the distance traveled $r(T)$, provides the impetus for moving forward, as without this term the model has no reason to move. This problem is solved across 4 phases (single stance: flat-foot, and toe-only contact; double stance: heel-toe, and toe–flat-foot contact) using the direct multiple-shooting method [8] implemented in the software package MUSCOD-II [9]. The vector of phase switching times $\underline{v}$ is free to vary during the solution process as is the total duration T. Note that no experimentally measured kinematics nor ground forces appear in the OCP which makes this approach well suited for predicting gait patterns.

2 Results & Discussion

The kinematics of the model agree best at the hip, moderately well at the knee, but are different from the experimental data at the ankle (Fig. 2A-C). The normal ground forces have a similar magnitude and timing as the experimental data, but have high frequency content that is not present in the experimental records (Fig. 2D). The pattern in kinematic error leads us to believe that the foot model is responsible for some of the differences between the model's gait and the experimental data. Despite these errors, we are encouraged by the similarity of the model's gait to the data given that we are using simplified muscles. We are presently working to construct a more accurate foot-ground contact model to improve the kinematics and ground forces of the model's gait.

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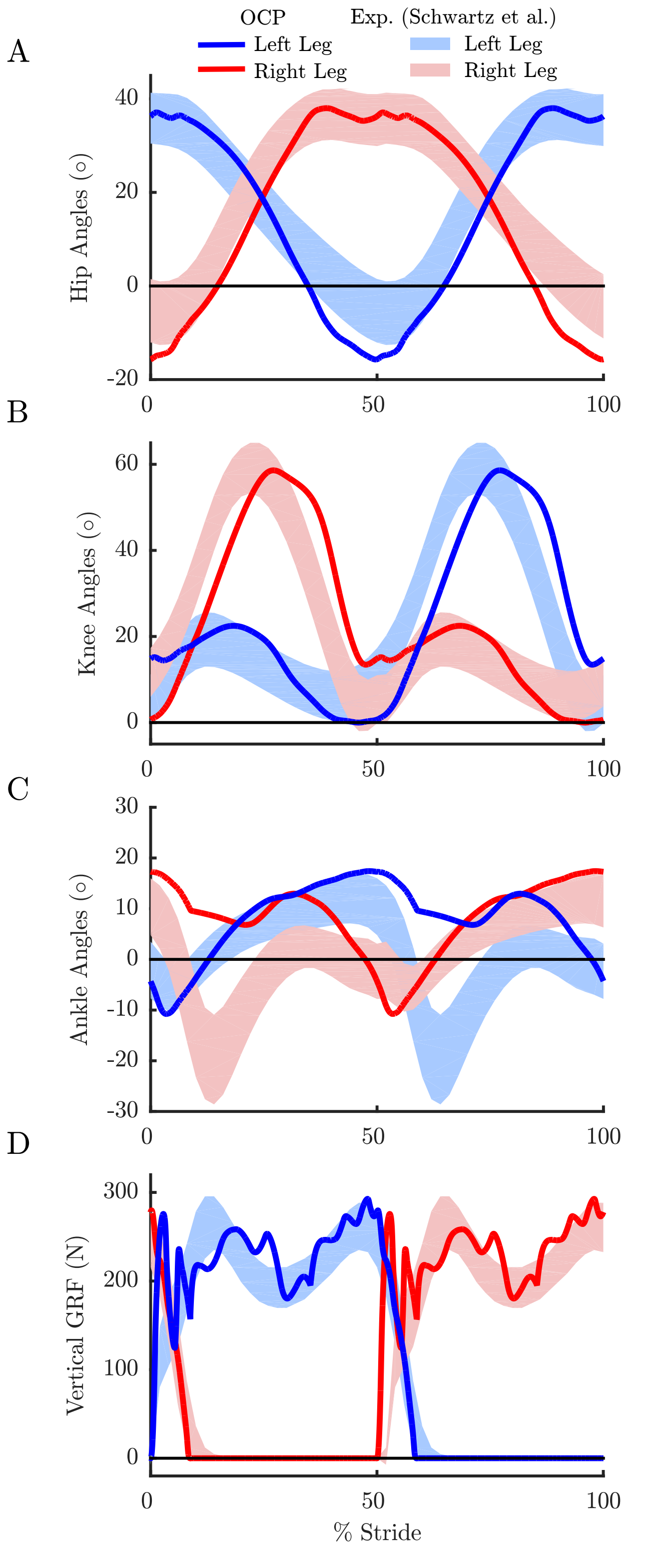

Figure 2: Kinematics and ground forces of the OCP solution for the left and right legs (blue and red lines respectively) plotted against normative data from Schwartz et al. [2].

References

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