ARE VIRTUAL MUSCLES AND REFLEX CONTROL CAPABLE OF DESCRIBING VARIATIONS WITHIN THE HUMAN GAIT CYCLE?

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INTRODUCTION

The design of modern prostheses and exoskeletons is focused on developing adaptive controllers which can more closely replicate able-bodied gait. Various control strategies have been proposed, though able-bodied locomotion is not usually achieved. It has been shown that virtual muscle models with autonomous reflex control can describe human locomotion and obtain realistic joint angles, torques, and muscle activations [1].

However, these reflex models have not been validated using human walking data obtained through experiments. Though reflex control may reproduce able-bodied walking, it may not be able to explain the variations within an actual human gait cycle. Here, we explore the possibility of using a reflex controller paired with virtual muscles, to determine if muscle reflexes alone are capable of reproducing the variability in human locomotion when subjected to random perturbations throughout the gait cycle.

METHODS

Walking data from 15 participants, including 4 females and 11 males, with an average age of 24 ± 4 years, height of 1.75 ± 0.09 m, mass of 74 ± 13 kg was used in the study [2]. The test subjects were perturbed using random belt acceleration signals generated from discrete-time Gaussian white noise. The variance of the signal was adjusted until the magnitude of the perturbations were within +/- 10% of the mean speeds of 0.8, 1.2, and 1.6 m/s.

However, randomly accelerating the treadmill belt, rollers, and motor will introduce inertial artifact errors in the sagittal plane moment of the ground reaction forces (GRF), which is used as an input in traditional inverse dynamics of human motion. The inertial errors in the sagittal plane moment were reduced by predicting the pitch moment from the belt acceleration using a linear, second-order, discrete-time model [3]. Joint angles and joint torques from the experiment were obtained through standard inverse 2D analysis [4], using joint positions obtained through the motion capture and the compensated GRF.

A planar, lower-leg model with three muscle groups (Gastrocnemius, Soleus, and Tibialis Anterior), representing a lower-limb prosthesis, was used to test the controller. Muscles were represented by a Hill-type model with a contractile element (CE) based on standard force-length and force-velocity properties, series/parallel nonlinear elastic elements (SEE/PEE), and a small amount of viscous damping in parallel to the contractile element.

Muscle contraction dynamics and activation dynamics were formulated as a set of first-order implicit differential equations (IDE) and were simulated in MATLAB® using a first-order, implicit Rosenbrock solver [5]. Predicted ankle torque (τ) can be obtained by multiplying the force generated by each muscle with the moment arms.

Muscle excitation signals (u) were obtained through an autonomous muscle reflex model using positive force feedback of the extensor muscles during the stance phase [1], where $S_{0,m}$, $G_m F_m$, $l_{OFF,m}$, and t_m are the pre-muscle stimulation, gained force, length offset, and time delay of each muscle, respectively:

Stance Reflexes:

$$u_{SOL} = S_{0,SOL} + G_{SOL}F_{SOL}(t - t_{SOL})$$

$$u_{TA} = S_{0,TA} + G_{TA}\left[\left(l_{CE,TA}(t_d) - l_{OFF,TA}\right)\right]G_{SOLTA}F_{SOL}(t - t_{TA})$$

$$u_{GAS} = S_{0,GAS} + G_{GAS}F_{GAS}(t - t_{GAS})$$

Swing Reflexes:

$u_{SOL} = S_{0,SOL}$

$u_{TA} = S_{0,TA} + G_{TA}[(l_{CE,TA}(t - t_{TA}) - l_{OFF,TA})]$

$u_{GAS} = S_{0,GAS}$

The control parameters were optimized in MATLAB using fminsearch, in which the objective function was to minimize the norm between the joint torque obtained through the experimental data and the estimated torque from the model. Muscle properties, including the SEE/PEE slack-lengths, maximum isometric force, and reflex time delays for each muscle were included as parameters in the optimization.

RESULTS AND DISCUSSION

The preliminary results of the optimization suggest that muscle reflexes can describe the majority of human locomotion, including some variability between gait cycles. Data from one test subject (male, age = 21, mass = 64 kg, speed = 1.2 m/s) is shown Figure 1, in which the experimental (black) and predicted (red) ankle torque are compared. The control parameters obtained through optimization, and the values previously described in literature [1], are shown in Table 1.

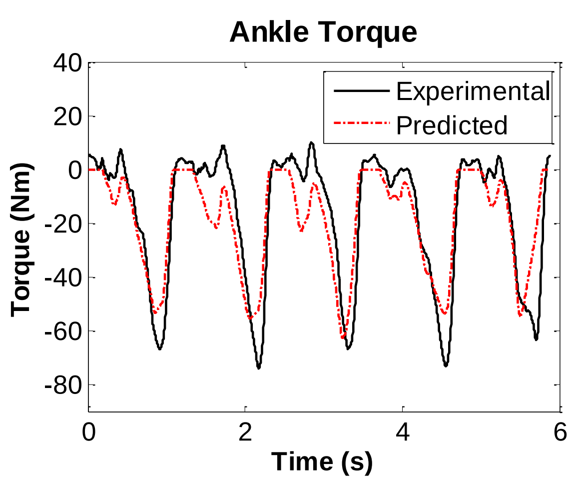

Figure 1: Experimental joint torque (black) and joint torque using optimized reflex control parameters (red)

A 11.03% difference between the torque signals was quantified by the root-mean-square error. Some of this error may reside in the remaining inertial artifacts that were not reduced during the compensation. However, the inability of the controller to reproduce the variations in the gait cycle cannot be explained by inertial errors alone. Though the controller produces a consistent magnitude of the torque signal, it does not account for the steps in which the human produced more torque to overcome the perturbation.

Although the presented results are not a complete study, these initial findings may suggest that reflex controllers may not be fully capable of replicating the variance in human locomotion, even after performing an optimization. Additional control, perhaps a model of a separate neurological process, could be added to muscle reflexes to better describe the disparities between gait cycles. Despite this limitation, current reflex controllers are sufficient enough for use in a powered prosthesis or exoskeleton, though their performance can be improved through further study.

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	G_{SOL}	G_{TA}	G_{GAS}	G_{SOLTA}	$S_{0,SOL}$	$S_{0,TA}$	$S_{0,GAS}$	$l_{OFF,TA}$	t_{SOL}(s)	t_{TA} (s)	t_{GAS} (s)
Literature	1.2	1.1	1.1	0.3	0.01	0.01	0.01	0.71	0.02	0.02	0.02
Optimization	2.1	0.26	0.1	0.26	0.0	0.01	0.02	0.55	0.03	0.04	0.02

Table 1: Control parameters obtained through previous literature [1] and through the results of the optimization