A real-time virtual muscle system for prosthesis control

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1 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 such as time-varying impedance control [1] have been proposed, though able-bodied gait is not usually achieved. We hypothesize that a more muscle-like actuation system, including the biarticular nature and nonlinear viscoelastic and dynamic properties of muscles, may simplify feedback control and improve performance. It has been shown through simulation that muscle properties contribute substantially to stability during walking when compared to torque-driven control [2].

Rather than build muscle-like hardware, we plan to use real-time simulation of muscle dynamics to control torques generated by electric motors. However, the differential equations that describe muscle dynamics are stiff and highly nonlinear. This will require small simulation time steps, making it difficult to achieve a simulation in real time. Recently, an implicit formulation of musculoskeletal dynamics was developed to achieve accurate simulations with minimal computation time [3]. Here, we explore the possibility of obtaining joint torques through a simulation of muscle dynamics, which would then be used to actuate the single-joint electric motors of an above-knee prosthetic leg.

2 Methods

A planar leg model with three monoarticular muscle groups (Vasti, Soleus, Tibialis Anterior) and three biarticular groups (Rectus Femoris, Hamstrings, Gastrocnemius) was used. System inputs were musculotendon length (L_m) derived from joint angles (q) from encoders at the hip, knee, and ankle and the neural excitation signals (u) for the six muscles, as shown in **Figure 1**. Knee and ankle torque $\tau(t)$ can be obtained by multiplying the force $F(t)$ generated by each muscle with the moment arms.

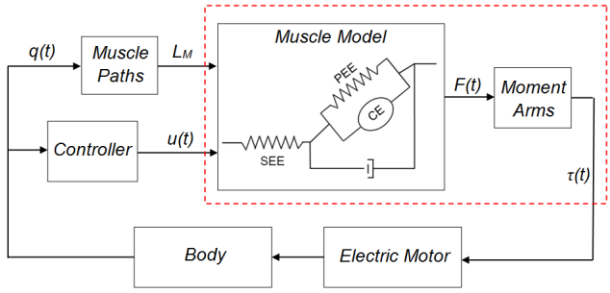

Figure 1. Block diagram of the muscle model, in which the red box indicates the scope of this current study

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 was formulated as a first-order implicit differential equation (IDE). An additional first-order IDE represented the activation dynamics. The muscle system therefore had 12 state variables, being the contractile element length and the activation state of the six muscles.

Two separate test cases were performed to verify that the muscle model produces reasonable muscle forces and joint torques. Initially, joint angles were held constant at zero while a step control fully actuated each muscle at different times throughout the simulation. The second condition used joint angle time histories *q(t)* for 30 seconds of normal walking and 1 Hz sinusoidal test signals for the muscle excitations *u(t)*. Simulations were performed in Matlab with a first order implicit Rosenbrock solver [3] using fixed time steps ranging from 0.08 to 16.0 ms. Numerical simulation errors were quantified as a percentage of the maximum joint torques $\tau(t)$ compared to the result from the smallest time step which was considered accurate. Computation times were measured in all tests.

3 Results

The muscle forces and joint torques obtained through the model responded appropriately to the prescribed muscle excitations. **Figure 2** demonstrates the first test case where joint angles were held constant at zero while a step control fully actuated each muscle. Predicted torques in the knee (blue) and ankle (green) correspond as expected to the muscle activations in the thigh and shank. The torque directions in the knee and ankle are influenced by muscle activations in the front and back of the leg. Additionally, the biarticular properties of muscle are represented when the gastrocnemius (cyan) is activated, in which both the knee and ankle torques are affected.

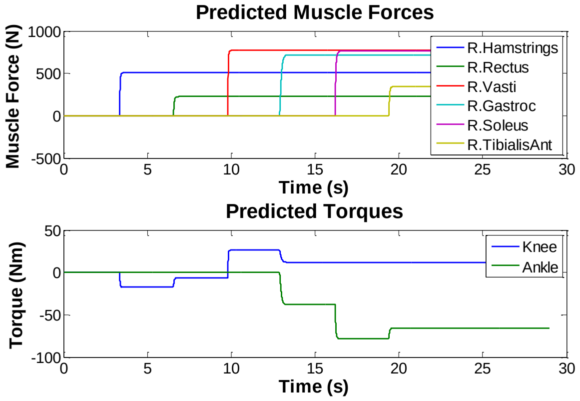

Figure 2. Predicted muscle forces (top) and torques (bottom) with constant joint angles *q(t)* of zero and step control *u*

The accuracy and speed of the simulation depended on the integrator step size, as shown in **Table 1**. Real-time is achieved at step sizes as small as 0.18 ms with simulation errors below 1%. Though larger time steps still produced acceptably accurate results, the model became unstable for time steps larger than 16 ms.

Simulation Step Size (ms)	**% RMS error Knee moment**	**% RMS error Ankle moment**	**Solution Time (s)**
0.10	0.14	0.07	40.80
0.18	0.25	0.13	23.10
1.60	1.44	0.71	2.50
16.00	15.48	6.75	0.26

Table 1: Integrator step size and its effect on accuracy and computation speed. Error was quantified as a percentage of the maximum moments in the knee and ankle. Simulation times below 30 seconds are faster than real time.

4 Conclusion

We have shown that torques can be obtained in a simulation of muscle dynamics with accurate results and minimal computation time. The implicit formulation of muscle dynamics and implicit integration method requires at each time step only one evaluation of the muscle dynamics and its derivatives, and solution of two state variables from two linear equations, for each muscle. This ensures that it can be used in an embedded system that uses low-level programming languages. The robustness of the control will be evaluated in simulation before implementing in an actual prosthesis or exoskeleton. Improvements to the model include the addition of feedback control that mimic reflexes from virtual muscle spindles and Golgi tendons.

References

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