Development of a Human Neuromuscular Balance Controller

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1 Introduction

Falls during locomotion are common in the elderly population, causing severe injuries such as fractures [1]. Fall prevention could be improved, for example by designing effective interventions, if human balance control is better understood. By simulating human movements, the effect of individual components of the human neuromuscular control system on balance can be investigated separately.

Our aim is to simulate human movements with a neuromuscular controller consisting of feedback pathways that are known to exist in humans. This abstract shows preliminary work where the neuromuscular controller was implemented in the hip. The knee and ankle were controlled by a position controller. The controller was able to stabilize a human model in a simulated perturbed standing experiment, and its behavior was compared to a human in the same experiment. We show that the motion in the hip correlated best with the motion of the human in the experiment.

2 Methods

The simulations used the sagittal plane human model with seven Hill-type muscles in each leg presented, which was in [2]. Symmetry was assumed in the control of the left and right leg. Multibody dynamics were simulated in Webots (Cyberbotics Ltd., Lausanne, Switzerland), while muscle dynamics were simulated in CasADi [3].

A neuromuscular controller was applied to the hip. Each muscle was connected to the spinal cord with two interneurons: one sending information on muscle length and one sending information on muscle force. Additionally, information on the ground reaction forces in the heel and toe assent to the spinal cord to simulate the cutaneous nerves. For each muscle, a motoneuron signal was defined using the following control law for muscle i:

$$\begin{aligned} u_i(t) = \quad & u_{0,i} + w_{pres,i}\frac{F_{y(toe)}(t-\tau_f)-F_{y(heel)}(t-\tau_f)}{F_{y(toe)}(t-\tau_f)+F_{y(heel)}(t-\tau_f)} \\ & + w_{len,i}(l_{CE,i}(t-\tau_i) - l_{CE(off),i}) \\ & + w_{for,i}\frac{F_{SEE,i}(t-\tau_i)}{F_{max}}, \end{aligned} \tag{1}$$

where u_0 is the basal activation, w_{pres} the pressure feedback gain, F_y the vertical ground reaction force, w_{len} is the length feedback gain, l_{CE} the normalized contractile element length, $l_{CE(off)}$ the normalized length threshold, w_{for}, the force feedback gain, F_{SEE} the force in the series elastic element, and F_{max} the maximum isometric force. τ_i denotes the time delay of muscle i, which was 5 ms for hip muscles. The time delay of the cutaneous feedback, τ_f, was 20 ms. The motoneuron signal was then sent to the muscles as muscle stimulation signal with another time delay of 5 ms.

The basal activation, length feedback gain, length threshold, force feedback gain, and pressure feedback gain were optimized using particle swarm optimization with a lexographic ordering extension to handle multiple objectives [2]: the centre of mass (CoM) and centre of pressure should be at a desired position inside the base of support, and activation and passive torque were minimized. This optimization was performed on a moving ground.

The knee joint and ankle joint were controlled using the position controller in Webots. The desired position was updated with a frequency of 1000 Hz. The desired knee angle was 10.7 degrees. The desired ankle angle was set such that the center of mass was in the middle of the foot.

Experimental data were recorded for a participant (height: 180 cm, weight: 79.12 kg) using an experimental protocol approved by the institutional review board of Cleveland State University (IRB-FY2018-40). The participant was asked to keep balance on a moving plate without stepping. Position data of 27 markers on the participant's body were recorded using a motion capture system with 10 Osprey cameras and Cortex software (Motion Analysis, Santa Rosa, CA). All data were filtered with a second order Butterworth filter with a cut-off frequency of 16 Hz [4]. Sagittal plane ankle, knee, and hip angles were calculated using markers on joint centres of rotation and the shoulder.

A sagittal plane perturbation signal was designed using random square waves with five amplitudes $[-5,-2.5,0,2.5,5]$ cm, and six durations $[0.25,0.5,0.75,1.0,1.25,1.5]$ s. Amplitudes and durations were randomly selected to generate a perturbation signal of 300s. A random 30 s sample was using in the simulation.

Joint angles were compared between the simulation and experiment. Correlation coefficients were calculated to determine how well the controller explained the observed motion in the experiment.

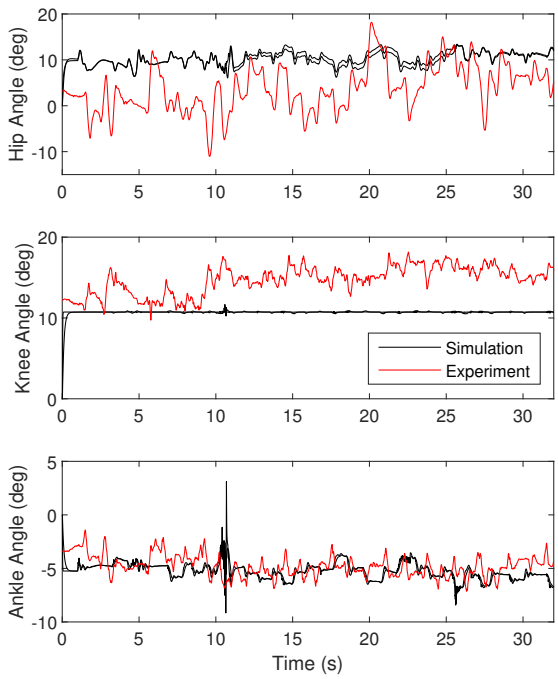

Figure 1: Joint angles, flexion positive, of the perturbed standing experiment and its simulation.

3 Results and Discussion

Figure 1 shows the simulated and experimental joint angles for the ankle, knee, and hip. The hip and knee range of motion were smaller in the simulation than in the experiment. The ankle range of motion was similar. The knee showed a bias with the experimental data after 10 s, while the hip joint showed a bias in the first 20 s. Correlation analysis revealed a correlation coefficient of 0.33 in the hip, 0.05 in the knee, and 0.08 in the ankle.

This means that the neuromuscular controlled could explain 33% of the motion in the hip. The correlation in this joint was higher than the correlation in the other joints, which were controlled by position control. This indicates that, as expected, the neuromuscular controller is more similar to the human control approach than a position controller.

The position controller in the knee could explain only 5% of the motion. Knee motion is often ignored in analyses of perturbed standing, and the human is modeled as in inverted pendulum with an ankle and a hip joint (e.g. [1]). Therefore, the knee position was fixed. However, the low correlation of the simulated knee motion with the experiment implies that there is knee motion in perturbed standing which should be taken into account in simulations.

An ideal position controller was applied to the knee and ankle joint. The position of the ankle was set to ensure that the CoM remained inside the base of support. This approach simplified the optimization of the controller parameters. Next, parameters for the neuromuscular controller will be optimized for the other joints.

4 Future Work

The balance system during gait might not be same as the balance system used during standing [5]. However, similar to this work balance is often studied in standing (e.g. [6]), or gait (e.g. [7]) separately.

Our aim is to create a simulation in which both gait and standing balance can be tested. Therefore, a simulation should be created of gait from start to finish, including the following phases: (1) balanced standing, (2) gait initiation, (3) steady-state gait, (4) gait termination and back to balanced standing. The work presented in this abstract was a first step towards phase 1 of this goal.

Next, a gait initiation will be designed (phase 2). This controller will consist of the following stages: calf muscle relaxation and activation of the tibialis anterior to shirt the CoM forward, and activation of the calf muscles in one leg to create push-off [1]. Thresholds, such as the CoM location, will define switches between stages. Steady-state gait in phase 3 will be controlled using reflexes and central pattern generators, similar to [2].

5 Conclusion

A neuromuscular controller with feedback on the muscle length, muscle force, and ground reaction force was implemented in the hip. In perturbed standing simulation, this controller explained 33% of the hip motion of an actual perturbed standing experiment.

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