The Role of Trunk Mechanics in Uphill and Downhill Walking: A Simulation and Experimental Study

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1 Introduction

The Spring-Loaded Inverted Pendulum (SLIP) model has been widely used to study bipedal locomotion, effectively capturing the fundamental dynamics of walking and running on level surfaces [1]. However, its constant energy level presents a challenge when modeling locomotion on slopes, where energy input is required for ascending and dissipation is necessary for descending.

Introducing a trunk into the SLIP model adds degrees of freedom, enabling energy adjustments through trunk dynamics. However, this also raises challenges in controlling trunk movement and hip torque. The virtual pivot point (VPP) concept and the force-modulated compliance (FMC) model have been proposed as effective strategies for trunk control on level ground [2,3], but their effectiveness in slope walking remains unexplored.

This study investigates whether incorporating a trunk into the SLIP model and VPP and FMC-based control strategies can facilitate locomotion on inclined terrains. By combining simulation and experimental studies, we assess how trunk dynamics contribute to energy modulation during uphill and downhill walking. Our findings improve our understanding of bipedal locomotion on slopes, providing insights to develop more accurate biomechanical models and improve assistive device design.

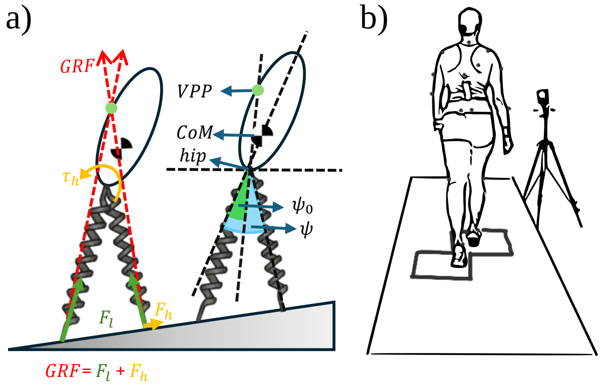

Figure 1: a) The SLIP model with a trunk walking on the slope. b) Experimental setup.

2 Methods

2.1 Simulation Methods

To investigate the role of trunk dynamics in walking on the slope, we extended the traditional 2D SLIP model by incorporating a trunk segment attached to the hip joint. To regulate the trunk's motion, we employed a FMC controller [2], a bioinspired approach that dynamically adjusts hip joint stiffness based on leg force feedback. This controller modulates the hip torque to maintain trunk stability. The FMC controller is designed to emulate the VPP hypothesis, which describes how ground reaction forces (GRFs) converge at a virtual point above the center of mass (CoM) in a reference frame attached to the CoM and aligned with upper body orientation [4,5]. This mechanism has been observed in human locomotion and is thought to contribute to natural and efficient gait patterns [2]. The hip torque generated by the FMC controller follows this equation:

$$\tau_h = cF_l(\psi_0 - \psi) \tag{1}$$

where τ_h represents the hip torque, c is a constant, F_l denotes the leg force, ψ is the angle between the upper body and the leg, and ψ_0 is the rest angle, defined as the angle between the upper body orientation and the hip-VPP line (see Fig. 1a).

To investigate the role of trunk dynamics in slope walking, we performed an optimization process using Bayesian optimization to determine the model parameters and initial conditions necessary for stable locomotion on a ±2° incline. After finding the solutions, the level walking was simulated with the same initial condition and model parameters (only ψ_0 set to zero).

2.2 Experimental Methods

For this study, nine able-bodied young adults walked up and down a 6 m long, instrumented ramp of 10° at self-selected speed (see Fig. 1b). There were three force plates (AMTI Optima BMS464508) integrated in the middle. Kinetic data were sampled at 1000 Hz. Kinematic data were measured with ten infrared cameras of Vicon (200 Hz). The subjects were prepared for 3D gait analysis following the standard protocol for the Plug-in Gait full body marker set.

For each condition (level walking, ramp up, ramp down), 12 valid trials were recorded.

3 Results

In the simulation, lowering ψ_0 shifts the VPP posteriorly, increasing forward trunk lean. A lower ψ_0 increases hip torque during the first half of the stance phase. Conversely, the strategy for downhill walking is the exact opposite. Increasing ψ_0 reduces hip torque in the first half of the stance phase (see Fig. 2a).

In human experiments, during the first half of the stance phase, the hip torque increased for uphill walking and decreased for downhill walking, as well (see Fig. 2b).

4 Discussion

Our results indicate that trunk dynamics and hip torque play a crucial role in generating the necessary energy for slope walking. Both in the simulation and the experiments, the hip torque increases in the first half of the stance phase for uphill walking and reduces for downhill walking, as shown in Figure 2.

In the model, decreasing ψ_0 in the FMC controller increases hip torque during the first half of the stance phase. This hip torque modulation produces a force perpendicular to the leg at the foot, reducing the braking component of the GRF and enabling uphill locomotion. The opposite occurs when increasing ψ_0 in the FMC controller. Increasing ψ_0 shifts the VPP backward and reduces hip torque in the first half of the stance phase. This reduced hip torque increases the braking component of the GRF, which can compensate for the energy added to the system due to down slope walking.

This mechanism is validated by comparing hip torque across different slopes. Although the hip strategy is not the only strategy humans use for walking uphill or downhill —other mechanisms such as the ankle mechanism at propulsion are also important [6]— our results show that the hip mechanism plays a significant role, especially in the first half of the stance phase, and the VPP concept can help predict this behavior.

The model's ability to generate gaits on higher slopes is restricted, indicating that while the FMC controller effectively regulates hip torque, it may not be sufficient on its own. To enable stable walking on steeper inclines, additional active components—such as variable damping or powered elements in the legs—may be required to provide the necessary force and adaptability [7].

The hip torque modulation strategy based on FMC has shown promise in designing assistive devices such as exoskeletons and exosuits for level-ground walking [8, 9]. Our results further highlight the potential of FMC-based controllers for extending this strategy to slope walking, indicating its applicability in more varied and challenging terrains [10].

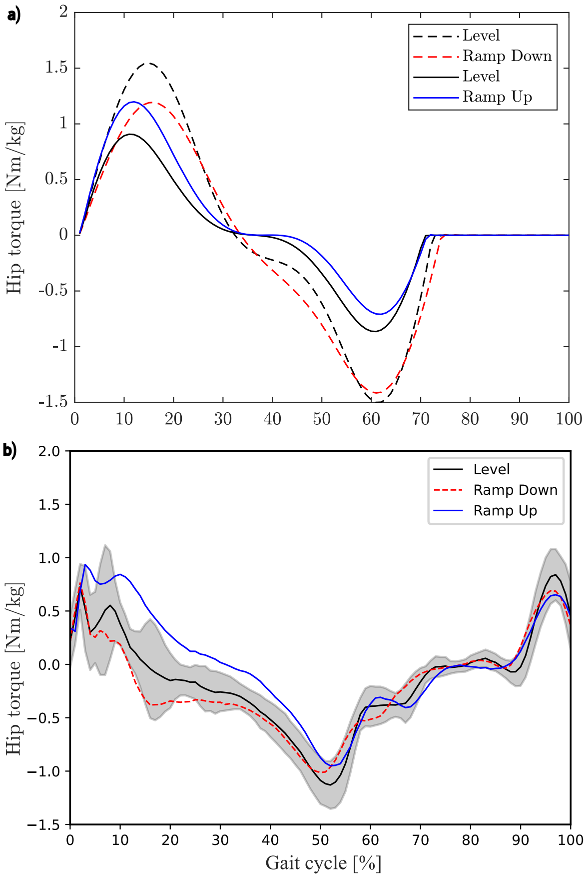

Figure 2: Hip torque. a) Simulated hip torque. We have two hip torque for level walking because the parameters for down slope and up slope gaits are optimized separately. b) Experimental hip torque (10° ramp). The gray shaded area is the standard deviation of level walking between participants.

5 Outlook

Next, we will investigate the contribution of different joints to slope walking through experiments and explore the active modulation of leg stiffness in the model [7] to further enhance our understanding of locomotion on inclined surfaces.

6 Acknowledgements

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