Ankle Controller Design to Enhance Sagittal Stability for Lower Limb Amputees

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1. Introduction

Lower limb amputees experience a 20 percent increase in metabolic energy consumption during normal walking [1], a disadvantage that can be reduced by active prostheses [2, 3]. Amputees also experience an increased rate of falls [4], a deficit that has not yet been addressed by active prostheses. Such falls might be due to decreased stability, suggesting that stabilizing prosthesis control could have a benefit. To study dynamic stability in amputees, we will use a limit-cycle based simulation approach. Similar approaches have been used to stabilize walking robots [5, 6, 7] and to explain fundamental aspects of able-bodied [8] and clinical [9] human gait. We propose to apply this technique to study amputee gait stability.

Joint actuation, in particular at the hip and ankle, may provide useful control inputs for stabilizing amputee gait. In robotics, hip actuation has received attention as an effective control input [5, 7]. Most lower-limb amputees maintain control of their hips, yet exhibit reduced stability. Perhaps differences in ankle actuation and sensing could help explain this deficit, while improvements to prosthetic ankle function could help restore balance. Both nominal [6] and step-to-step variations of [6, 7] ankle actuation parameters influence stability in limit cycle robots. We will investigate these ideas in a simple model with powered hip and ankle joints and compare the influence of different control inputs on stability. If ankle control has a significant effect compared to hip control, we might enhance stability for amputees with prosthetic ankles.

The control of robotic assistive devices must address several challenging problems not faced by walking robots or non-amputee humans, such as limited state information and asymmetric ankle actuation. A robotic ankle can gain only local ankle information and contains states unavailable to the amputee's biological control system. Amputees also have asymmetric ankle actuation and asymmetric gaits overall [10], which may affect stability. We will model these features and study their effects.

The purpose of this study was to explore controllers that improve stability for lower-limb amputees. We developed a limit-cycle model of amputee gait, focused on ankle actuation, and studied the effects of nominal ankle actuation parameters and various step-to-step LQR control designs on stability. We compared the effectiveness of ankle and hip actuation strategies, and investigated state estimation to compensate for data inaccessible to either the human or the prosthesis.

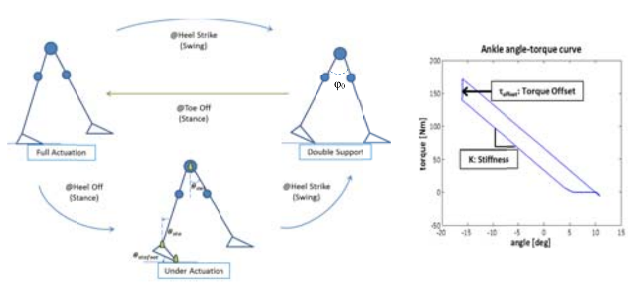
2. Methods and Results

We developed a limit cycle walking model (A), and used it to study the effects of ankle actuation (B) and the influence of asymmetric walking (C) on stability. We evaluated stability using random disturbances to floor height during walking.

A. Model

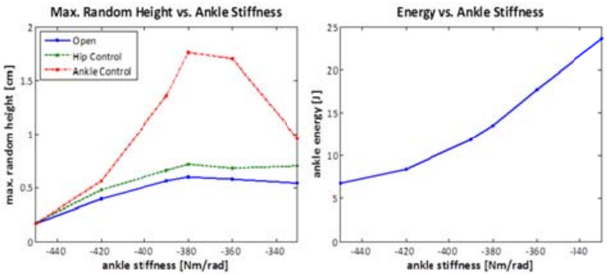
We developed a simple model of walking to investigate the effect of ankle actuation on stability (Figure 1a) and found limit cycles with features similar to human gait. The ankle was actuated by mimicking the ankle-torque relationship of able-bodied humans [11]. This relationship was approximated by a piecewise linear curve with two parameters, ankle torque offset and ankle stiffness (Figure 1b). Hip actuation was modeled as either a Spring Hip, with a low-stiffness spring and no damping, or a PD Hip, with critically-damped proportional derivative control to keep fixed step length [5]. Various limit cycles were found with identical speed (1.25 $m \cdot s^{-1}$), and step length (0.7 m).

We also designed step-to-step state feedback controllers using LQR. At the end of each step, model states were compared to desired states, control inputs were calculated, and changes applied in the next step. We used this technique to modulate ankle torque offset, nominal hip angle, and/or hip stiffness.



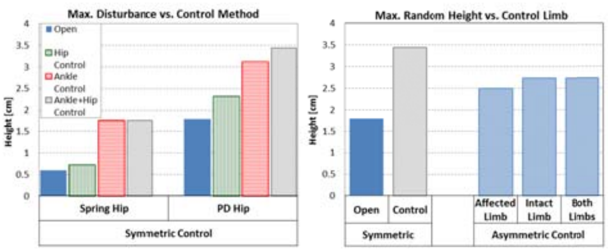
(a) (b)

Figure 1: Model. (a) Dynamic model: this under-actuated model shows angle representation with a lumped mass at the hip and on each leg. φ_0 is nominal hip angle. The walker also shows three different continuous dynamics, and the arrows represent transitions. From the full actuation phase, the walker can go to either double support or under actuation. The decision is based on either the heel strike event of the swing leg or heel off of the stance leg. When the walker is in the under actuation phase, it eventually arrives to double support phase once swing leg hits the ground. Then the walker comes back to the full actuation phase by going through the toe off event (stance leg). (b) Ankle angle-torque curve: τ_{offset} - constant torque offset. K-stiffness. Dorsiflexion ankle torque is calculated as K times ankle angle, and the plantar flexion ankle torque is calculated as K times ankle angle plus τ_{offset}.



(a) (b)

Figure 2: Parameter study. (a) Stability measure vs. ankle parameters. Open: open loop. Hip Control: hip stiffness step-to-step control. Ankle Control: ankle torque offset step-to-step control. Different ankle stiffness presents different stability. The results suggest that ankle torque offset control is the most effective to improve stability. (b) Energy input vs. ankle parameters.



(a) (b)

Figure 3: Stability measures. (a) Comparing of spring hip and PD controlled hip (fixed step length). Open means there is no step to step control. Hip Control: spring hip – hip stiffness control, PD hip – nominal hip angle control. Ankle Control: ankle torque offset control. Ankle + Hip Control: both ankle torque offset and hip control. Maximum random height plot shows that ankle control provides increased stability with/without step-to-step hip control. (b) Stability result considering limited state information. PD hip is used. Left plot shows open loop walker (open), and ankle and hip controlled walker (control) using full state information. In right plot, affected limb controlled walker (affected limb), intact limb controlled walker (intact limb), and both limbs controlled walker (both limb) are shown. For the asymmetric limb control, full states are estimated using available states. The results show that with state estimation, the robotic ankle can increase walking stability compared to open.

B. Effects of ankle actuation on stability

We measured stability and energy use while varying ankle stiffness in the Spring Hip model (Figure 2), and found that open-loop stability was strongly affected, with an optimum near the net stiffness observed in human walking [11]. Use of step-to-step feedback control at the hip or ankle increased disturbance rejection, but did not alter this trend. Ankle stiffness influenced premature heel rise angles and collision timing [12] which in turn seem to have affected stability.

We also compared several hip and ankle control strategies, and found that ankle control always had a significant impact. We compared peak disturbance rejection using ankle-only, hip-only, and ankle-plus-hip state feedback control in both the Spring Hip and PD Hip models, all at the same speed, step length, and heel rise angle (Figure 3a). PD Hip control and step-to-step control of hip parameters both improved stability, consistent with prior results [5]. Yet even in the PD Hip model, step-to-step modulation of the ankle torque offset parameter alone nearly doubled disturbance rejection capacity, twice the effect of hip control alone. This suggests that appropriate step-to-step modulation of prosthetic ankle torques could lead to improved walking balance for amputees.

C. Asymmetric ankle actuation effects on stability

Robotic prostheses may only have direct access to measures of its own states, making the state feedback control strategy used here more difficult. The intact limb loses information from the amputated joint and surrounding tissues. To enable realistic implementation of LQR, we designed state estimators using a Kalman filter. The robotic ankle on the affected limb estimated the full state of the system using ankle states and calculated control inputs using the same gain matrices calculated previously. The intact limb used stance and swing leg information to estimate the full state, and also, separately, performed control actions at the intact hip and ankle joints. We then tested three combinations of controls: robotic ankle control, intact limb control, and simultaneous, but separate, control of both limbs. Figure 3b shows that even if only the robotic ankle is controlled, stability can be increased significantly compared to open loop. Lower stability of controlling both limbs using state estimation shows that information loss might be a fundamental cause of the reduced stability observed in amputees. Even though the effects of prosthetic ankle control in concert with amputee control were modest, there could be benefits due to reduced control effort.

3. Open questions

Will the effects of ankle control be as significant in a 3D model of walking? Can our stabilizing controllers be made even more robust by non-linear control techniques, for instance mapping to the limit cycle from states outside the linear region? How significant are the tradeoffs between energy consumption and stability? Can the present results be usefully applied to real assistive devices? Let us discuss.

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