Effect of Environmental Factors on Level of Trip Disturbance: A Simulation Study

Ming Liu, *Member, IEEE*, Peter Bohlen, and He (Helen) Huang, *Senior Member, IEEE*

Abstract— **Above knee amputees exhibit a higher risk of falling than able-bodied people, so the capacity to recover from trips (a major cause of unintentional falls) is critical for these amputees to prevent fall-related injuries. Although trip recovery approaches using powered prostheses have been proposed, the effectiveness of these approaches has not been evaluated with varied trip-related disturbance levels. Here, we conducted a simulation study to understand the relationship between trip-related disturbance levels and environmental factors. This knowledge could clarify the design space as well as guide design and evaluation techniques of future trip recovery approaches.**

I. Introduction

Compared to able-bodied people, lower limb amputees, especially above knee amputees, are at high risk of falling [1]. To prevent fall-related injuries and mortality [2-5], current research efforts aim to build safer environments for amputees and train amputees to protect themselves during unexpected falls. The recent development of powered prostheses decreases amputees' chance of falling by generating large foot clearance during swing phase to avoid trips [6] that account for more than 50% of unexpected falls [7].

Despite these efforts, unexpected disturbances (e.g. trips) cannot be avoided completely. Intensive engineering efforts have been conducted to design advanced prostheses to help amputees recover from trips using the two strategies used by able-bodied people: elevating strategy and lowering strategy. Current, commercially-available knee prostheses permit amputees to use the lowering strategy [8, 9], but their capacity to conduct the elevating strategy is still under development.

Current researches focus on two topics related to trip recovery: 1) early detection of trip events and 2) design trip recovery approaches for powered prosthetic legs to recover from trips. Lawson et. al. developed a trip detector based on accelerometers embedded on the prosthetic legs [10]. Zhang et. al. demonstrated the capacity to detect trips in amputees using a combination of residual muscle EMG and mechanical signals [11]. Thatte et. al. proposed a hierarchical swing-leg placement controller [12]. In the higher level, the control goal is set to track desired foot placement location; in the lower level, knee-joint torque is calculated to generate specified, phase-dependent knee kinematics. Based on our knowledge, this approach has not been demonstrated on amputees yet. Another interesting trip recovery controller has been demonstrated on transfemoral amputees, which first detects a trip and then implements a recovery approach [13, 14].

Most of the existing research is based on experimental data collected in well-controlled lab environments, using only one type of obstacle, which may not represent the variety of levels of disturbances in amputees' everyday life that lead to falls. An understanding of varied trip disturbance levels will help researchers determine a design space for trip recovery systems. In turn, it could guide the design and evaluation techniques of efficient recovery systems, highlighting the reliability and effectiveness of these recovery systems. These are important techniques to establish especially when clean trip tests are very hard to conduct.

Two types of factors are known to affect the level of disturbance: environmental factors and internal factors. The environmental factors include material properties of the obstacle and foot and obstacle location. The location of the obstacle affects the relative speed between the toe and obstacle when the trip happens. The material properties of the foot and obstacle further affect trip durations and relative speeds between the toes and obstacles when the trips end. The internal factors are lower-limb dynamic properties [15].

It is challenging to study the relationship between environmental factors and the level of disturbance experimentally. As shown in [16-19], subjects will modify their gait patterns to improve their chances of recovery if they expect a trip. These anticipatory actions alter dynamic properties of the lower limb joints and easily contaminate experimental results if one subject has to be tripped more than once. Very few studies have been conducted to understand the level of disturbance involved in different trip scenarios. One study is based on a single trip case using a rigid aluminum plate as the obstacle [20]. The disturbance related with trips was usually represented by a fixed impulse at the toe in both simulation and experimental studies [12, 21] .

In this study, we conducted a forward dynamics simulation to understand the relationship between the level of trip disturbance and environmental factors. Besides measuring disturbance, which was quantified by the change of joint angle and angular velocity caused by trips, we also considered the chance that the foot contacted the ground during a trip or just after a trip. These contacts may also contribute to the trip disturbance by generating a large friction force on the foot. The results of this study may lead to a clear design space for trip recovery approaches in assistive devices such as powered prosthetic legs to restore the user's balance after trips. An enhanced capacity to recover from trips will improve amputees' mobility and stability in everyday life.

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M. Liu, P. Bohlen and H. Huang are with the NCSU/UNC Joint Department of Biomedical Engineering, NC State University, Raleigh, NC, 27695; University of North Carolina at Chapel Hill, Chapel Hill, NC 27599.

Corresponding author: Ming Liu (e-mail: mliu10@ncsu.edu, phone: 919-515-8541).

II. METHODS

We used a simplified dynamic model to understand the interaction between the foot and obstacle in the sagittal plane. The model was built with SimMechanics, a Simulink toolbox (MathWork, MA, US), and included two sub-models: a rigid-body model for the locomotion simulation and a nonlinear spring-damper model for the tripping simulation.

A. *Rigid Model for Locomotion*

A simplified rigid-body model was constructed to represent the motion of the leg experiencing tripping. To mimic the locomotion of above-knee amputees, who usually rely on fixed, passive ankle joints, the thigh and shank-foot are modeled as rigid bodies. The motion of the leg was described by coordinates of the hip center, hip joint angle, and knee-joint angle. The inertial properties of each segment were determined from the height and weight of an able-bodied subject using the formulas reported in [22]. The interaction force acted on the toe of the foot during the tripping.

B. *Nonlinear Spring-Damper Model for Tripping*

In order to estimate the level of disturbance generated during a trip, a simple nonlinear spring-damper model was integrated to simulate the collision between the foot and obstacle [23]. An obstacle was defined as a block fixed on the ground with one of its surfaces perpendicular to the ground. The normal forces (F_{normal}), which were paralleled with the ground, were calculated using the following non-linear spring damper expression from [23]:

$$F_{normal} = kx^e + Step(x, 0, 0, d_{max}, c_{max})\dot{x} \qquad (1)$$

where x is the distance traveled by the toe, and $\dot{x}$ is speed of the toe. This equation is further explained in [23]. Frictional forces ($F_{friction}$) during the collision were simulated as Coulomb friction and calculated using:

$$F_{friction} = \mu F_{normal} * sgn(\dot{y}) \qquad (2)$$

where $\dot{y}$ is the vertical velocity of the toe, and μ is the coefficient of friction. The coefficient of friction is a static coefficient (μ_{static}) when vertical velocity is less than 0.05 m/s and dynamic ($\mu_{dynamic}$) when the velocity is greater than or equal to 0.05 m/s.

Although more sophisticated collision models are available [24], this model was chosen due to its simple structure and wide acceptance. The same model has been used in commercial rigid-body simulation software Adams for collision simulations.

C. *Simulation Procedure*

A forward dynamic simulation was used to illustrate the influence of environmental factors on the level of disturbance (Fig. 1). This simulation was conducted based on kinematic and kinetic data collected from an able-bodied subject (weight: 80 kg, height: 1.74 m) during treadmill walking at a self-selected walking speed. The protocol was approved by the IRB at the University of North Carolina at Chapel Hill. The subject was given informed consent before the experiment. A video-based motion capture system (VICON, Oxford, UK) collected the motion data and we calculated the torque at different joints using inverse dynamics.

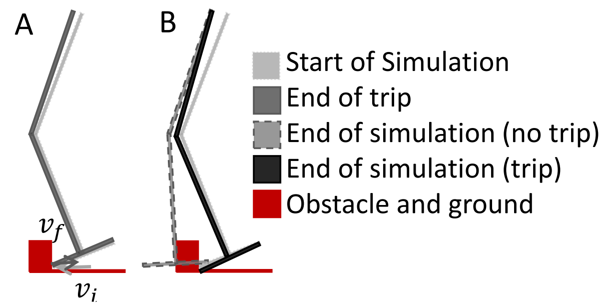

Figure 1. An example case of the forward dynamics simulation. A: before collision and after collision; B: Influence of the collision at the end of the simulation.

The elastic properties between the foot and obstacle were defined by 6 parameters (Table 1), which were adjusted in the simulation. We used 5 potential values evenly spaced between the minimum and maximum values of each parameter as shown in Table 1. The range of each parameter was chosen based on values reported in [25] and further expanded to cover more elastic and inelastic scenarios.

TABLE I. PARAMETER VALUE RANGES

Variable	***Min Value***	***Max Value***
Spring Coefficient (k in N/m)	$1x10^4$	$2x10^5$
Spring Exponent (e)	100	1500
Max Damping Coefficient (c_{max} in N/(ms))	1	2.2
Max Damping Penetration (d_{max} in m)	.001	.02
Static Friction Coefficient ($\boldsymbol{\mu_{static}}$)	.2	.8
Dynamic Friction Coefficient ($\mu_{dynamic}$)	.1	.2

To align with existing studies, which rely on the trip onset time to specify trip events, we defined the locations of the obstacles based on the trip onset time (as a percentage of swing time) instead of the distance between the obstacle and a reference point (i.e. ipsilateral toe at the preceding toe off). Because the toe moved forward during swing, the high percentage of swing phase indicated that the obstacle was far from the reference point.

The tripping events at 15%, 30%, 45%, 60%, and 75% of swing phase were studied in the simulation. We used the measured joint angle and angular velocity at the hip and knee joints at the selected the trip time as initial conditions for the forward simulation. The obstacle was set 0.01 mm ahead of the toe at the selected trip time.

Several assumptions were adopted to simplify the model. First, joint torques were constant during the simulation. Because trips were never anticipated, it was reasonable to assume that the torque acting on each joint was not altered during or just after the trip. It was only altered after the subjects reacted to the trip through reflex and/or voluntary reaction. Here, the torque on each joint was calculated based on inverse dynamics at the time of trip using the synchronized motion data. Second, the trajectory of the hip center was not affected by the trip. Third, the contact surface of the obstacle was assumed to be perpendicular to the ground. Fourth, the potential of foot-ground contact during the trip and shortly after the trip was ignored.

The simulation duration was 100ms to fully illustrate the environmental factors' influence. Based on the data reported in [16], reflex actions were observed on average 80ms after a trip as indicated by EMG signals on able-bodied subjects. Considering the 20ms electromechanical delay commonly seen in the lower limb [26], we concluded that in the first 100 ms after a trip, there was no correction torque from human subjects.

D. Results Presentation

The level of disturbance of the trip was presented as the change of angle and angular velocity at the hip and knee joints at the end of the simulation compared with measured non-collision data.

Because multiple parameters were involved in defining the material properties of the foot and obstacle, it was hard to interpret the influence of each parameter independently. Here, we used coefficient of restitution (COR), a well-defined physical concept, to generalize the material properties. COR is a measure of the relative velocity between two bodies (A and B) before and after they collide and is calculated by:

$$COR = \frac{v_{Af}-v_{Bf}}{v_{Bi}-v_{Ai}} \qquad (3)$$

where v_{Af} and v_{Bf} are the velocity of object A and object B after the collision, respectively, and v_{Ai} and v_{Bi} are the velocity of object A and object B before the collision, respectively. COR is often used to describe the elasticity between the two objects that collided. When the COR equals one, the collision is elastic and no energy is lost during the collision. When the COR equals zero, the collision is inelastic and the two objects move together. Because the obstacle was fixed in our simulation, the COR was calculated based on toe velocity before and after the trip.

III. Results

The simulation results showed that the level of disturbance was heavily affected by the environmental factors. Fig. 2 illustrates the level of disturbance in relation to obstacle locations and material properties of the foot and obstacle. The disturbances at the knee-joint were linearly related to the material elasticity, which aligned with the definition of COR. The disturbance at the hip joint was dominated by obstacle locations.

Additional knee flexion and hip extension were observed, which was caused by the trip. This observation aligned with the fact that the interaction force between the foot and the obstacle was against the direction of locomotion, which would cause knee flexion and hip extension.

To validate the simulation results, disturbance observed in a tripping test conducted by the Providence VA Medical Center on one human subject was also included in Fig. 2. The trips happened around 32% of swing phase with a rigid plastic plate. This subject used to validate our simulation and the subject used to build it had similar heights and weights. Due to the data acquisition frequency limitation of the motion capture system (150Hz), the exact COR could not be determined, so the level of disturbances (100ms after the tripping) were represented by a straight line between COR=0.45 and COR=0.65, which was the expected COR range for a collision between the subject's foot and a hard plastic plate. We can see this representation in Fig. 2(a) and (b) where the disturbance in this tripping case (red line) intersects the disturbance at 30% swing phase (dashed line). These intersections indicate good agreement between the model-predicted disturbance and observed disturbance at the knee joint. Although no intersection was seen in Fig. 2(c), the red line and the dash line (30% swing) are close to each other, which also indicates a good model prediction accuracy. The only expectation was the Fig. 2(D), where the observed level of disturbance was far from the real disturbance level, which could have been caused by subject expecting a trip during the experiment and compensating more quickly (i.e. during the 100ms after the trip).

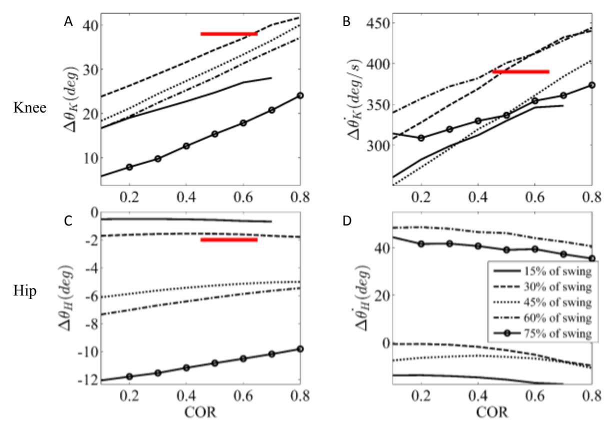

Figure 2. Average level of disturbance caused by tripping at the end of the simulation. A depicted the change in the knee angle $\Delta\theta_K$; B depicted changes of hip angle, $\Delta\theta_H$; C depicted change of knee angular velocity, $\Delta\dot{\theta}_K$; and D depicted change of hip angular velocity, $\Delta\dot{\theta}_H$. Flexion was positive for both knee and hip joints. The red lines were measured disturbance on a human subject during single tripping case. In D, the measured disturbance was about 180 degree/sec.

Fig. 3 shows the average minimum foot clearance from the ground during the simulation. During mid-swing, negative foot clearance was observed, which indicates a risk of uncontrolled contact between the foot and ground.

Although the simulation results generated cases where COR was smaller than 0.1 and larger than 0.8, these cases were ignored because such cases were not commonly seen. COR smaller than 0.1 indicated the foot was almost stuck on the obstacle, and COR larger than 0.8 were only reported for highly elastic collisions, such as ping-pong balls bouncing on a hard wood floor.

IV. Discussion

Our simulation results clearly identified the fact that the level of disturbance caused by the trip was affected by environmental factors. Because trips were never predicted and related environmental factors were all unknown, a recovery activity must consider all the different level of disturbance to ensure the reliability of recovery.

During the elevating strategy, flexion at the ipsilateral knee-joint and hip-joint was necessary to generate enough foot clearance to avoid a second collision with the obstacle. The extension disturbance on the hip joint, especially at the mid-to-late swing, made it harder to generate the necessary clearance. An additional hip motion would be required to ensure the success of the trip recovery if the elevation strategy was used.

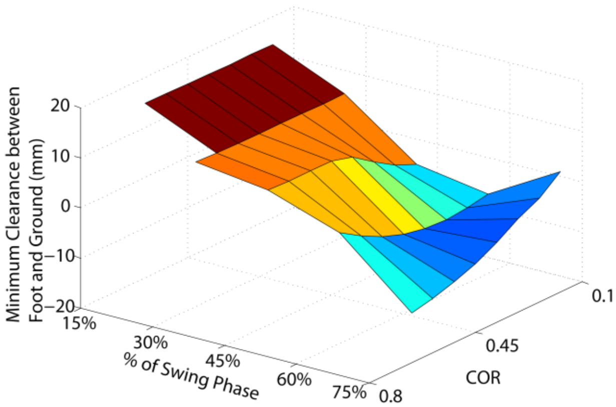

Figure 3. Average minimum foot clearance during the simulation.

Although we did not simulate the contact between the foot and ground during the simulation, the uncontrolled contact would make the elevating strategy even harder to apply because the needed knee flexion would be damped by the friction force during contact. The chance of uncontrolled contact might be decreased by providing some dorsiflexion during the swing phase, which is commonly seen in able-bodied people.

The uncontrolled foot-ground contact also could be used to explain the inconsistency in selecting a recovery strategy (i.e. elevating, lowering) at mid- swing in addition to the involved mechanical work as reported in [27]. The uncontrolled contact might eliminate the chance of recovery using the elevating strategy and make the lowering strategy mandatory.

Limitations of this study include 1) the motion data were collected on a single able-bodied subject, and 2) the model was only validated using one trip case. We will test the consistency of our results by involving more cases in future studies. We are also looking into novel approaches to simulate human reactions to trips. A validation procedure could be developed to test the efficiency of trip recovery approaches without heavy involvement of human subjects.

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