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ORIGINAL ARTICLE

Predictive simulations of elevating and lowering strategies in human stumble recovery

Tökezleme sonrası toparlama sürecinde uygulanan yükseltme ve alçaltma stratejilerinin öngörücü benzetimleri

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Abstract

Purpose: Older adults and individuals with neuromuscular impairments face a high risk of falls, which can be mitigated by identifying effective stumble recovery strategies for rehabilitation. Studying stumble recovery through empirical methods is challenging due to injury risks and constraints on natural movement, whereas predictive neuromechanical simulations offer a viable alternative. This study aimed to use a musculoskeletal model within a predictive simulation framework to analyze human stumble recovery following anterior directed perturbations.

Methods: Using a simplified musculoskeletal model and a reflex-based neural controller, two different scenarios for perturbations occurring in the early (20%) and late (60%) swing phases were simulated. The kinematics of the swing leg, including hip, knee, and ankle joint angles, were analyzed for similarity to real human stumble recovery. Additionally, recovery strategies were identified by tracking the swing leg's toe trajectory following perturbation.

Results: Early swing perturbations elicited an elevating strategy, increasing hip and knee flexion to clear the obstacle, while late swing perturbations triggered a lowering strategy, rapidly placing the foot to restore stability. Minor deviations from experimental data were observed, particularly in ankle dorsiflexion and swing phase duration.

Conclusion: This study highlights the effectiveness of predictive neuromechanical simulations in analyzing stumble recovery. The framework successfully replicated key recovery mechanisms, demonstrating its potential for rehabilitation, assistive device design, and fall prevention strategies aimed at enhancing mobility and reducing injury risk in vulnerable populations.

Keywords: Stumbling, Simulation, Falls, Biomechanics.

Öz

Amaç: Düşme riski, yaşlılar ve dengeyi etkileyen nöromüsküler bozuklukları olan hastalar başta olmak üzere pek çok bireyin karşı karşıya olduğu bir durumdur. Tökezleme sonrası etkili toparlama stratejilerinin rehabilitasyon programlarına dahil edilmesi ile bu risk azaltılabilir. Ancak bu stratejilerin deneysel yöntemlerle incelenmesi, yaralanma riski ve harekette ortaya çıkabilecek kısıtlılıklar nedeniyle zordur. Bu zorlukları gidermek için, bu çalışma, öngörücü nöromekanik simülasyonlar kullanarak insanların anterior yönlü pertürbasyonlar sonrası ürettiği toparlanma hareketini analiz etmeyi amaçlamaktadır.

Yöntem: Basitleştirilmiş bir kas-iskelet modeli ve refleks tabanlı bir sinirsel denetleyici kullanılarak, erken (%20) ve geç (%60) salınım fazlarında meydana gelen pertürbasyonlara yönelik iki ayrı senaryonun simülasyonu gerçekleştirildi. Salınım fazındaki kalça, diz ve ayak bileği bilek eklem açıları tökezleme sonrası kurtarma hareketiyle benzerlik açısından analiz edildi. Ayrıca, pertürbasyonun ardından salınım fazındaki bacağın ayak parmağının izlediği yörünge takip edilerek modelin kullandığı kurtarma stratejileri belirlendi.

Bulgular: Erken salınım fazında uygulanan pertürbasyon, engeli aşmak için kalça ve diz fleksiyonunda artış ile karakterize bir yükseltme stratejisi ortaya çıkarırken, geç salınım fazında uygulanan pertürbasyon, dengeyi yeniden sağlamak amacıyla salınım fazındaki ayağın hızla yere indirilmesi ile karakterize bir alçaltma stratejisini tetikledi. Özellikle ayak bileği dorsifleksiyonunda ve salınım fazı süresinde deneysel verilerden küçük sapmalar gözlendi.

Sonuç: Bu çalışma, öngörücü nöromekanik simülasyonların tökezleme sonrası doğal kurtarma hareketini analiz etmedeki etkinliğini vurgulamaktadır. Gerçekleştirilen simülasyonlar, tökezleme sonrası ana toparlanma mekanizmalarını başarılı bir şekilde taklit etmiştir. Öngörücü benzetimlerle elde edilen verilerin rehabilitasyon programlarının geliştirilmesinde, yardımcı cihaz tasarımlarında ve mobiliteyi artırarak yaralanma riskini azaltmayı amaçlayan düşmeyi önleyici stratejilerin geliştirilmesinde önemli bir potansiyele sahip olduğunu göstermektedir.

Anahtar Kelimeler: Tökezleme, Simülasyon, Düşme, Biyomekanik.

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INTRODUCTION

Falls are a major public health concern, particularly among older adults, who experience them at a significantly higher rate due to age-related declines in physical, sensory and cognitive functions.[1] While the risk of falls is commonly associated with the elderly, an increased susceptibility to falls is observed for other vulnerable populations such as individuals with neurological disorders including Parkinson's disease and dementia,[2] as well as stroke survivors.[3] Considering number of populations associated with high risk, the consequences of falls impose a substantial burden on healthcare systems, leading to a range of injuries such as fracture,[4] traumatic brain injuries,[5] and long-term mobility impairments.[6] These injuries contribute to extended hospital stays, rehabilitation costs, and increased mortality rates.[7] Research indicates that a significant proportion of falls can be prevented through early identification of risk factors, balance training, and assistive interventions.[8,9] From a biomechanical perspective, falls are defined as the sudden, unintended loss of balance that results in an individual making unintended contact with the ground or a lower surface.[10] This loss of balance often arises due to an inability to adequately recover from a destabilizing perturbation, such as a stumble A stumble is defined as a loss of balance caused by an unexpected disturbance, such as tripping over an obstacle or encountering an uneven surface, without necessarily resulting in a fall. While all falls involve a failure to regain balance, not all stumbles lead to falls, as individuals can often recover stability through neuromuscular responses. Hence, understanding the biomechanics of stumbles and the neuromuscular mechanisms underlying balance recovery is therefore critical for developing effective fall prevention strategies, ultimately reducing both individual suffering and healthcare expenditures associated with fall-related injuries.[11]

Stumble recovery involves a sequence of rapid neuromechanical responses aimed at restoring gait stability following an external perturbation. In this study, we define "stumble" as a disruption in normal walking, which can occur through various mechanisms, including foot scuffing, tripping over obstacles, or slipping on low-friction surfaces. Our focus is specifically on stumble recovery following an anteriorly directed perturbation, induced by the sudden obstruction of the swing foot by an external obstacle. This type of perturbation is particularly relevant in understanding the dynamics of stumbling during everyday activities, such as walking in crowded environments or navigating uneven terrain, where the foot may unexpectedly collide with an obstacle in front of the body. Human stumble recovery strategies are highly phase-dependent, with individuals employing different responses depending on the timing of the perturbation within the swing phase of the gait cycle. Early swing phase stumbles typically activate the elevating strategy, while late-phase stumbles trigger the lowering strategy[12,13.]

Studying stumble recovery through empirical methods is challenging due to ethical concerns and the risk of injury, resulting in limited direct data collection. Computational approaches, such as predictive simulations, provide a valuable alternative by modeling biomechanical responses to balance perturbations in a controlled, risk-free environment. These simulations integrate biomechanics, motor control, and optimization algorithms to predict human movement, offering insights into joint loading, muscle coordination, and movement strategies.[14] They not only have the potential to enhance our understanding of stumble recovery but also aid in developing interventions to improve stability and prevent falls. In rehabilitation, predictive simulations help model patient-specific movement patterns, identifying dysfunctions and informing personalized treatment plans.[15] By analyzing muscle activation and joint dynamics, they guide targeted rehabilitation exercises for individuals with impaired balance and assess the potential impact of interventions like physical therapy or surgery before implementation.[16] Additionally, they assist in designing gait-support devices, such as lower limb prostheses.[17] Within this framework, predictive simulations have the potential to address the limitations associated with experimental data collection, particularly in scenarios where direct measurement is impractical or ethically constrained.

This study aims to develop a predictive simulation framework for simulating human-like stumble recovery following an anteriorly directed perturbation caused by the swing foot colliding with a rigid obstacle. Specifically, it focuses on simulating stumbles occurring during the early and late phases of the swing cycle, at 20% and 60% of the swing phase, respectively, to capture the phase-dependent mechanisms involved in stumble recovery. Additionally, the study introduces a framework for analyzing these recovery responses, with implications for improving gait stability and informing the design of rehabilitation and assistive devices. Ultimately, this work contributes to advancing our understanding of balance recovery during walking and the prevention of fall-related injuries.

METHODS

Participants

The simulation framework for this study were built on a validated predictive model for normal human gait developed in Veerkamp et al.[18] This framework was adopted as the foundation for our study, with modifications to simulate stumbling perturbations. The details of their simulation framework pertinent to our study are as follows and for further details on the musculoskeletal model and neural controller used in this study, readers are referred to their original work:[18]

The base framework was implemented using SCONE, an open-source predictive neuromechanical simulation software.[19] SCONE specializes in modeling and analyzing human movement dynamics, employing advanced optimization algorithms to predict movements that optimize biomechanical objectives like energy efficiency and stability. It integrates seamlessly with tools like OpenSim, providing insights into muscle activations and joint mechanics.[20]

The musculoskeletal model utilized in this study was derived from OpenSim's gait2392 model, a validated computational framework for human biomechanics.[21] Originally comprising 23 degrees of freedom to represent major joints such as the hip, knee, ankle, and lumbar spine, the model was refined to enhance computational efficiency. The motion was constrained to the sagittal plane, and the trunk and pelvis were merged into a single rigid body, reducing the models degree of freedom to nine. Although it could be argued that combining the head, arms, and trunk into a single unit may overlook certain dynamic interactions, this simplification is well justified. It is established that paraspinal muscle activation has a minimal relationship with trunk kinematics during stumble recovery. Furthermore, forward deceleration of trunk flexion is predominantly managed through passive control mechanisms or the action of hip extensors.[22]

The 92 original muscle-tendon units, modeled using Hill-type dynamics, were lumped into 18 units to streamline optimization. The feet were represented with two Hunt-Crossley contact spheres per foot, with parameters estimated through tracking simulations in SCONE. These modifications preserved essential biomechanical details, enabling the model to accurately simulate muscle activations and their contributions to joint torques during locomotion.

The neural controller used in this study is a widely recognized model for replicating muscle coordination in normal gait.[23] It uses neural feedback loops and muscle activations to simulate adaptive motor responses, mirroring human motor control during dynamic activities like walking.

The physiologically-based objective functions utilized in the optimization process include cost of transport, muscle activation, head acceleration, and ground reaction force jerk. These functions correspond to walking effort, muscle fatigue, head stability, and injury risk, respectively.

To replicate the typical progression of a stumble event followed by successful recovery (Figure 1) in predictive simulations, the optimization process begins just prior to the impact and continues across successive gait cycles until the model achieves stable locomotion. Two optimization scenarios were conducted to simulate perturbations during the early and late swing phases, corresponding to 20% and 60% of the swing phase, respectively. These phases were selected based on the available experimental data in the literature,[12,13] which primarily focuses on the early and late swing phases. Although SCONE allows for the optimization of initial model states, the initial conditions were not optimized deliberately to prevent the model from pre-adapting to the impending perturbations. Instead, the initial states, comprising positions, velocities, and muscle activations, for both scenarios were derived from the predictive simulation results for normal gait reported in Veerkamp et al.[18] Figure 2 presents the model states immediately prior to impact for these two scenarios.

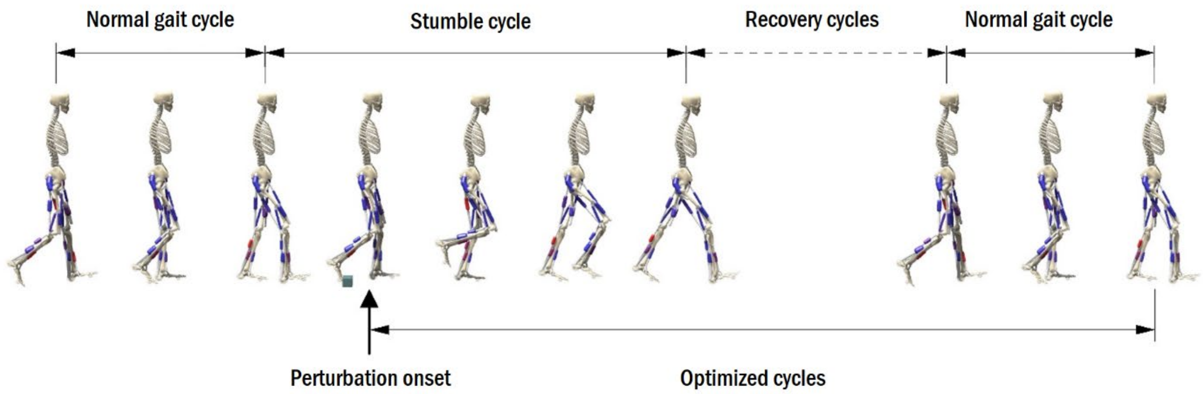

Figure 1. Typical progression of a stumble caused by obstruction of the swing foot.

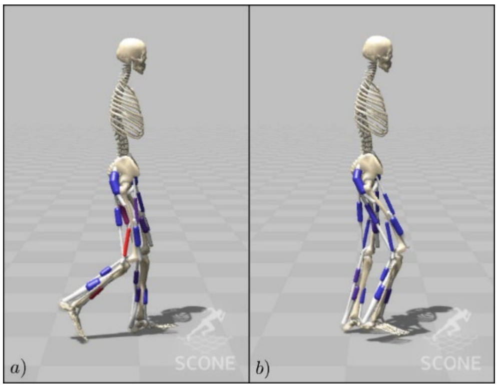

Figure 2. Initial conditions prior to stumble at a) early swing phase (20% into the swing phase) and b) late swing phase (60% into the swing phase)

To accurately replicate the perturbation caused by the obstruction of the swing foot, a constant anteriorly directed force of 100 N was applied to the tip of the model's right toe at early swing phase (20% into the swing phase) and late swing phase (60% into the swing phase) for a duration of 0.10 seconds. The magnitude and duration of the anteriorly-directed perturbation were estimated based on values reported in the literature,[24,25] where the vertical component of perturbation was found to be negligible. Therefore, we focused solely on an anterior perturbation force in our optimization.

In the optimization process, the primary objective was to minimize the cost of transport (CoT), a commonly used metric for evaluating locomotor efficiency.[18] The cost of transport quantifies the energy expenditure required for movement relative to the distance traveled and the model's mass. It was computed as:

$$CoT = \frac{1}{distance \times mass} \times \int_0^{t_{end}} \sum_{m=1}^{18} \dot{E}_m(t)\, dt$$

where $\dot{E}_m(t)$ represents the instantaneous energy consumption of the m^{th} muscle at time t, integrated over the total simulation duration t_{end}. To ensure the model has sufficient time to achieve stable locomotion, t_{end} was set to 10 seconds.

The simulation results were first evaluated based on whether the model successfully recovered from perturbation. In this context, a successful stumble recovery in the simulation was defined based on the stability and continuity of locomotion. A fall is considered to occur if the model's center of gravity drops below a predefined threshold, at which point the simulation terminated. Recovery is considered successful only if the model achieves stable locomotion by the end of the simulation (t=10 s). Stability is quantified by assessing the similarity of consecutive gait cycles, with recovery deemed successful if the kinematics of successive cycles exhibit an R^2 value above 0.950.

For simulations that achieve successful stumble recovery, further analysis was conducted to examine the strategy selection of the model. This was done by tracking the trajectory of the swing leg's toe throughout the perturbed gait cycle. We also quantified step length and step duration during the stumble cycle. Once distinct recovery strategies were identified, the kinematics of the swing leg were compared with normative gait data derived from predictive simulations of healthy gait in Veerkamp et al.,[18] which also served as the initial conditions for this study. Additionally, the recovery strategies selected by the model were qualitatively compared with experimental findings from the literature to assess their biomechanical plausibility.

RESULTS

The optimization of both early and late swing phase stumbles was successful, with each scenario achieving a stable gait cycle by the end of the simulation. The simulation framework effectively generated human-like stumble recovery motions (Figure 3-5). In the early swing phase, the swing foot followed the elevating strategy by lifting the swing leg directly over the obstacle without additional steps after impact. In the late swing phase, recovery motion mirrored the human-like lowering strategy. Unlike the early swing phase, the swing foot was placed in front of the obstacle shortly after the impact, while obstacle clearance was initiated by the contralateral (support) leg.

The elevating strategy resulted in a step length of 0.75 m and a step duration of 0.70 s during the stumble cycle. Compared to the normative values of 0.65 m and 0.56 s, respectively, this indicates an increase in both step length and duration. In contrast, the lowering strategy led to a step length of 0.21 m and a step duration of 0.44 s, showing a substantial reduction relative to normative gait. In the elevating strategy, the maximum hip flexion angle reached 47°, compared to the normative value of 33°, while the maximum knee flexion angle was 100°, exceeding the normative value of 59° (Figure 4). Conversely, in the lowering strategy, knee flexion at heel strike was 41°, higher than the normative value of 4°, while ankle dorsiflexion at heel strike was 5°, lower than the normative value of 14° (Figure 5).

DISCUSSION

In this study, we aimed to develop a framework for predictive simulations of human gait stumble recovery to overcome the limitations of empirical data collection. We achieved this by modifying an existing predictive simulation framework for healthy gait to account for perturbations caused by foot-obstacle contact. Specifically, we introduced a perturbation force at the moment of impact to simulate the destabilizing effect of stumbling. Additionally, unlike the original framework, we did not optimize the initial states of the simulation. Instead, we used pre-existing predictive simulation data of healthy gait to set the initial conditions at 20% and 60% of the swing phase, corresponding to early and late stumble events, respectively. The simulation results for both the early and late swing phases effectively captured the phase-dependent nature of the recovery motion. Previous studies indicated transition from the elevating strategy to the lowering strategy happened at an average of 44% into the swing phase.[13] In our simulations, the framework generated an elevating strategy for stumbles occurring in the early swing phase (at 20% into the swing phase) and a lowering strategy for those occurring in the late swing phase (at 60% into the swing phase). This demonstrates that the simulation framework successfully replicates the overall characteristics of the recovery response in both phases of the gait.

The main characteristics of elevating strategy involves over flexion of hip and knee

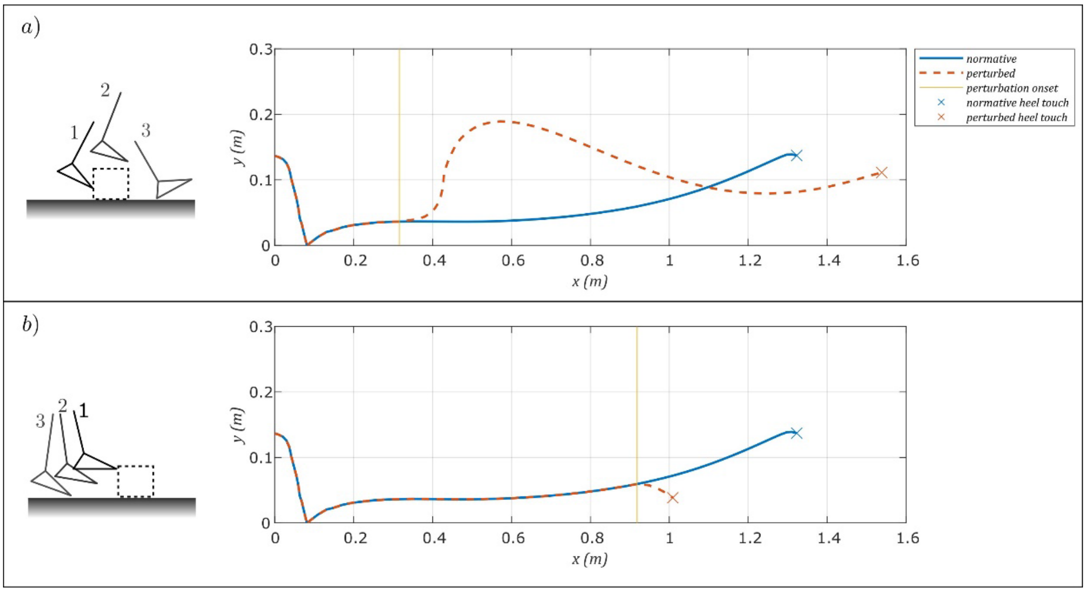

Figure 3. Schematic illustration (adapted from Shirota et al.27) and toe trajectory, shown in orange, of the swing leg for perturbations applied in the early swing phase (a) and late swing phase (b). The normative toe trajectory without any stumble is shown in blue for both perturbations.

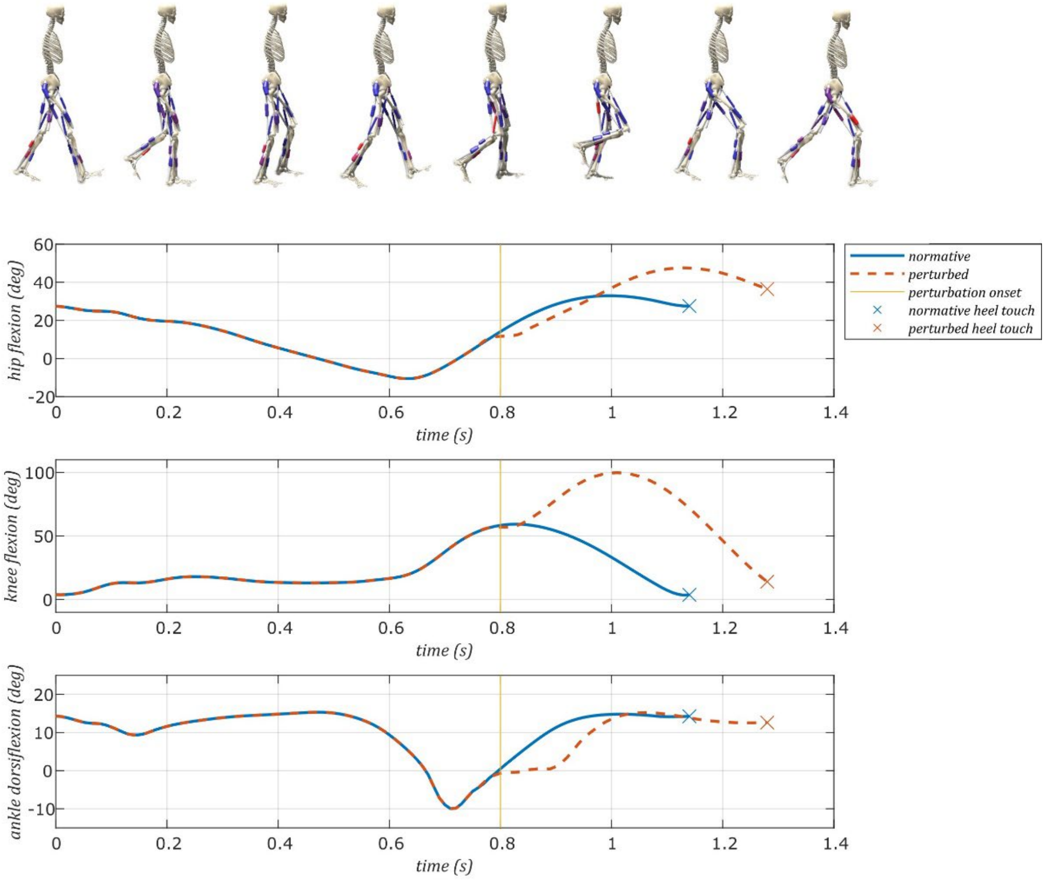

Figure 4. Sagittal plane kinematics for normative (blue) and perturbed (orange) gait. The perturbation is applied in the early swing phase (20% of the gait cycle).

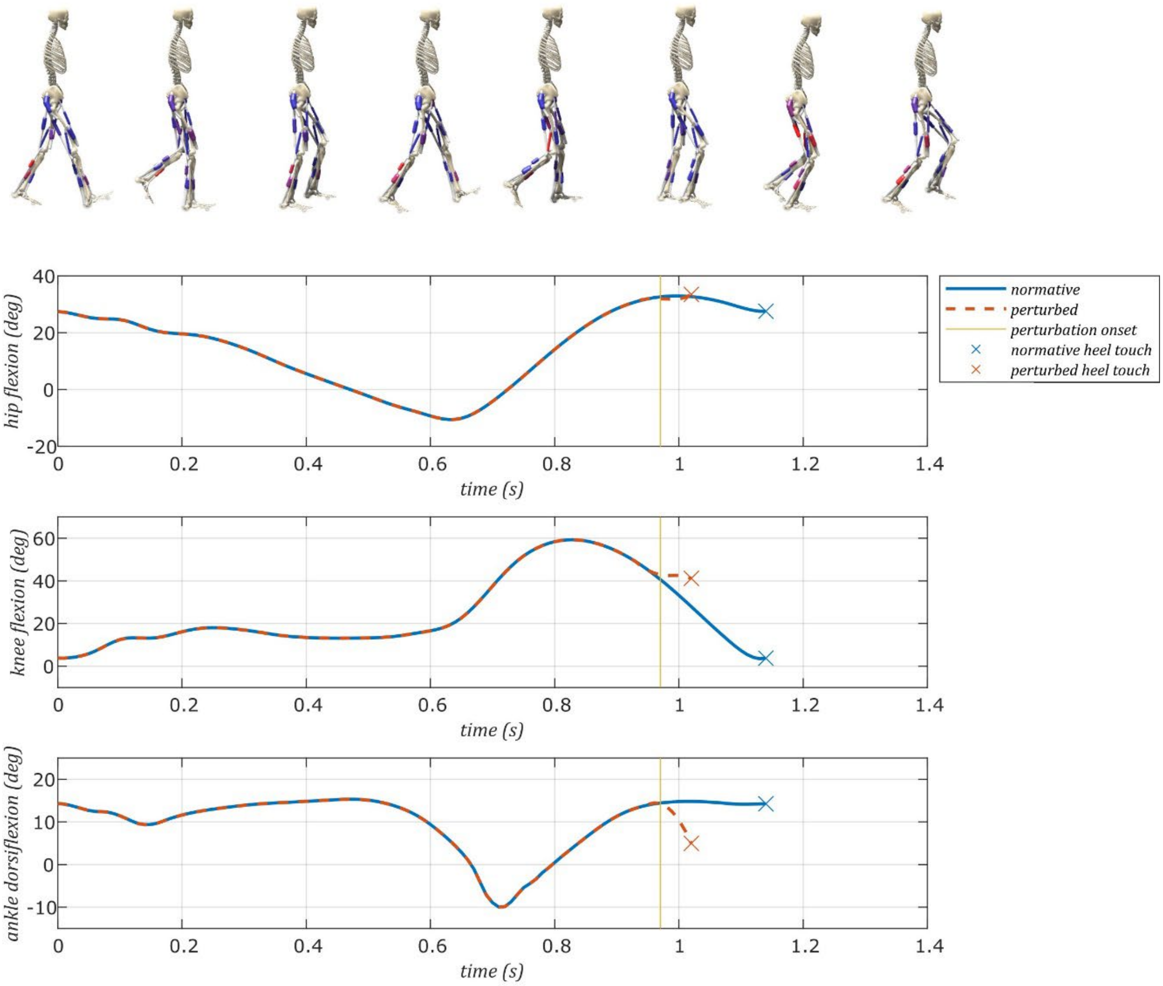

Figure 5. Sagittal plane kinematics for normative (blue) and perturbed (orange) gait. The perturbation is applied in the late swing phase (60% of the gait cycle).

joints in an attempt to lift the swing limb over the obstacle after the stumble. In the early swing phase simulation, a similar pattern is captured in these joints. As a result, over flexion of hip and knee joints enabled the model to perform the elevating strategy. However, it was observed that the ankle joint could partially capture the human like motion. Similar to the experimental data reported in Schillings et al.,[13] right after the stumble the ankle is plantar flexed due to the perturbation force. After the foot is cleared from the obstacle, compared to the normative data, ankle joint is over dorsiflexed to prepare for the heel strike. In simulation, we can see that right after the perturbation force is applied the ankle joint is plantar flexed however, in subsequent section the ankle dorsiflexion remained level with the normative data.

In the lowering strategy, human movement patterns differ markedly from those observed in the elevating strategy. Following perturbation, the subsequent knee extension and ankle dorsiflexion responses are inhibited, leading to either a flattened heel contact or a forefoot landing, accompanied by increased knee flexion at the moment of landing. The movement outcomes of the lowering strategy in the late swing simulation indicate that the model exhibits forefoot landing, consistent with experimental data reported in Eng et al.[12] Furthermore, the model demonstrates minimal variation in hip flexion. Similar to the experimental findings, the knee extension and dorsiflexion responses following impact are inhibited, and foot lowering is primarily achieved through ankle plantarflexion. The perturbation at this phase prompts the model to react more rapidly than anticipated. Experimental data suggest that the lowering strategy results in a shorter swing phase compared to the elevating strategy, although it remains longer than that observed in normative data. However, in the simulation, the model produces a significantly shorter swing phase, failing to exhibit the increased knee near heel strike.

The interaction between the swing foot and the obstacle can be characterized as an impact event, which occurs over a relatively short duration and generates substantial reaction forces.[26] Due to the impulsive nature of such events, the position variables remain continuous, while the velocities experience an instantaneous discontinuity. Consequently, a temporal delay is required for the velocity change to manifest in the positional variables: hip flexion, knee flexion, and ankle dorsiflexion. This phenomenon is also evident in the experimental data reported in.[12,13] However, in both strategies resulting from predictive simulation, joint angle variations commence immediately following the perturbation onset. This discrepancy represents a notable deviation between the simulation results and the experimental observations.

These findings are not only theoretically consistent with experimental studies but also hold strong clinical implications. The ability to simulate stumble recovery strategies with predictive accuracy supports the development of rehabilitation protocols tailored to specific phases of gait. For instance, training programs could be designed to strengthen hip and knee flexion for early-swing recovery (elevating strategy), or improve rapid foot placement and load acceptance in late swing (lowering strategy). Furthermore, the insights into altered ankle and knee mechanics during perturbation can inform the design of phase-sensitive wearable assistive devices such as smart prostheses, which must respond quickly and differently based on the gait phase to prevent a fall. By identifying neuromechanical deficiencies, such as limited ankle dorsiflexion or insufficient swing leg clearance, clinicians can target specific impairments that compromise balance recovery.

Limitations and future work

The presented approach has some limitations. First, we assumed a constant impact force throughout the swing phase, disregarding its variation depending on the velocity of the swing foot at perturbation onset. Phase-dependent formulation of the impact force should be implemented in future works. Additionally, our analysis primarily focused on the kinematics of the swing leg, as it plays a central role in strategy selection. However, the stance leg also contributes significantly to recovery dynamics, particularly in redistributing body weight and stabilizing postural control. Future studies should incorporate a more comprehensive evaluation of stance leg mechanics to fully capture the interplay between both limbs during stumble recovery. Furthermore, the current model excluded upper limb dynamics by modelling the upper body as a single rigid body. Although this enhances the computational efficiency, the exclusion of upper limb dynamics may limit the accuracy of the simulated recovery strategies, particularly in capturing whole-body coordination. Future simulations would benefit from explicitly modeling the arms to better reflect their stabilizing role during balance recovery. Lastly, the scope of this study is limited to immediate neuromechanical responses to assess the ability of our framework to generate stumble recovery motion. However, the full stabilization process includes not only the initial corrective movement, but also subsequent adaptations in limb coordination, posture, and center of mass control. Future studies could expand the current framework to capture this extended recovery sequence. Modeling and analyzing the evolution of gait stability across multiple steps would offer a more comprehensive understanding of stumble recovery strategies and help identify subtle impairments in individuals at risk of falling.

Conclusion

This study successfully employed predictive neuromechanical simulations to model human stumble recovery strategies in response to anteriorly directed perturbations, providing valuable insights into the phase-dependent nature of these strategies. The simulation replicated key biomechanical features, demonstrating that perturbations occurring during the early swing phase predominantly trigger an elevating strategy, while those during the late swing phase elicit a lowering strategy.[12] The elevating strategy, characteristic of the early swing phase, involves lifting the perturbed limb to clear the obstacle, facilitated by increased hip flexion and knee extension, with the contralateral leg offering support.[13] In contrast, the lowering strategy, typical of the late swing phase, involves a rapid downward movement of the perturbed limb to make immediate ground contact, stabilizing gait through increased knee flexion and rapid foot placement.[13] While the model accurately captured these kinematic characteristics, certain discrepancies, such as deviations in ankle dorsiflexion and swing phase duration, suggest areas for refinement.

These findings underscore the potential of predictive simulations for studying fall prevention and rehabilitation, particularly for populations at heightened risk of falls. By highlighting the adaptability of human gait through phase-dependent recovery strategies, this research emphasizes the critical role of neuromechanical control in maintaining stability during perturbations. Future improvements, such as enhanced impact modeling and a more thorough evaluation of stance leg contributions, will further increase the accuracy and applicability of these simulations. Ultimately, by bridging the gap between computational modeling and real-world biomechanics, this study lays the foundation for developing targeted interventions to improve stability and reduce fall-related injuries. In clinical contexts, such simulations can guide the personalization of fall-prevention interventions, particularly for older adults and individuals with compromised neuromuscular function. The phase-dependent strategies identified here can serve as benchmarks for assessing patient recovery potential and for training compensatory behaviors. Additionally, rehabilitation robotics and prosthetic control systems can be better informed by these neuromechanical patterns, enabling more adaptive and responsive support during gait perturbations.

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