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Deteriorated stabilization of walking in individuals with spastic cerebral palsy revealed by a simulated tripping perturbation

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Abstract

The aim of this study was to make use of a previously introduced method of a simulated tripping perturbation to examine stability of walking in individuals with and without cerebral palsy. This tripping perturbation is a forward-dynamics analysis, and it works so that in a subject-specific muscle-driven simulation of walking, created from experimental gait data, a force is applied to the swing-foot, and resulting changes in the kinematics are observed. Here, changes in the fore-aft position of the trunk were analyzed. Subjects were 10 children with spastic diplegic cerebral palsy and eight unimpaired children walking at their self-selected speed. Several tripping perturbations throughout the swing phase were performed, and each perturbation was used to analyze stability at a respective instant of time. At a given instant of time, walking was defined as stable if after initially deviating from its unperturbed position, because of the perturbation, the trunk then approached and stayed close to that position. Walking was in turn defined as unstable if the trunk moved away from its unperturbed position. All unimpaired subjects were stable at some point of their swing phases, whereas six out of the ten CP subjects were never stable. The unimpaired subjects were statistically significantly stable for a larger percentage of the swing phase than the CP subjects ($p < 0.05$). On the average, unimpaired subjects were stable for 8.5% of the swing phase while CP subjects were stable for 1.5% of the swing phase. These findings demonstrate that a tripping perturbation is capable of detecting a balance impairment. Thus, it shows promise as a tool for rehabilitation.

Background

Cerebral palsy (CP) results in a balance impairment observable in walking [1] [2] [3]. Over time CP deteriorates the walking function [4]. Thus, means of improving the walking function and stability are called for. An accurate method to assess stability is essential for effective rehabilitation. Such method can be utilized when searching for individuals with high fall risk, and when monitoring and evaluating outcomes of therapy or surgery in rehabilitation. It is particularly crucial to evaluate stability against falling. Falls are mostly related to tripping or slipping [5] [6], so stability should be evaluated under these circumstances. Yet, traditional measures of walking stability do not address stability under these circumstances. Therefore, stability with respect to tripping was investigated here.

Studying trunk kinematics is the key to understanding stability against falling. Ability to limit trunk motion after a trip has been shown to distinguish non-fallers from fallers [7]. Moreover, stability of the trunk during walking has been reported to be associated with fall history [8]. Control of the trunk is physiologically related to control of the overall balance such that it reduces the acceleration of head, which allows stabilization of the optic flow, a more effective processing of the vestibular-system signals, and a consequent control of balance [9] [10]. From a mechanical point of view, the trunk is massive and is located high in the body, so it stores a large amount of gravitational potential energy. Thus, the trunk must be controlled in order to avert transformation of potential energy to kinetic energy, i.e., falling. Control of the whole-body center of mass or angular momentum, for instance, does not guarantee such physiological and mechanical aspects of balance control.

Balance can be controlled in a proactive and reactive manner [11]. A reactive control means that the central nervous system reacts to a perturbation in order to minimize the impact of this perturbation. A proactive control means in turn that the central nervous system makes walking dynamics stable against a possible perturbation before and without being aware of the impending perturbation.

This study focused on the stability of walking that results from a proactive control. One way to test such stability is to apply a short perturbation to probe the mechanical state of the system. In this study, probing was carried out by a recently introduced method of simulated tripping perturbation [12]. This tripping perturbation is a forward-dynamics analysis, and it works so that in a subject-specific muscle-driven simulation of walking, created from experimental gait data, a force is applied to the swing-foot, and resulting changes in the kinematics are observed. Advantages of this tripping perturbation are that it is standardized for all conditions and subjects, ensuring their fair comparison, and that it can be done and repeated free of any anticipation or adaptation by the subject. The tripping perturbation is also ideal for studying a proactive control because it avoids the influence of the reactive control.

The aim of this study was to examine the stability of walking in individuals with and without CP, making use of a tripping perturbation. The following sections describe how this tripping perturbation was used to examine stability,

and report an observed difference between the subject groups with respect to stabilization of walking against tripping in the fore-aft direction.

Methods

The basis of the present study was in the simulations of Liu et al. [13] and Steele et al. [14], who created their simulations from experimental gait data so as to examine their subjects' gait. Those simulations are described in Section 2.1 below. In this study, these subjects were examined further by analyzing the simulations with a tripping perturbation [12] as described in Section 2.2.

Musculoskeletal model and simulation

Liu et al. [13] and Steele et al. [14] created subject-specific simulations of walking using the OpenSim software [15]. Both the software and simulations are freely available at www.simtk.org. The simulations were for children with spastic diplegic cerebral palsy [14] and for unimpaired children [13] walking at their self-selected speed. Age, body mass, and leg length of ten CP subjects were 8.1 ± 1.7 yrs, 27.1 ± 9.1 kg, and 0.65 ± 0.06 m, respectively [14]. For comparison, eight subjects – age 12.9 ± 3.3 yrs, body mass 51.8 ± 19.2 kg, leg length 0.81 ± 0.09 m, respectively – were studied [13]. Experimental gait data included kinematics, ground reaction forces, and electromyographic (EMG) recordings [14] [16].

The procedures for creating and testing the simulations have previously been described in detail [13] [14]. Briefly, a generic musculoskeletal model [17] [18] with 23 degrees of freedom and 92 muscle-tendon actuators, was scaled to match each subject's anthropometry. Subtalar and metatarsophalangeal joints were locked at neutral anatomical angles. Dynamic inconsistency between the measured ground-reaction-force data and the model kinematics was reduced with the residual reduction algorithm by applying residual forces and torques to the pelvis of the model, and adjusting the model's mass properties and kinematics [15]. Computed muscle control was employed for determining the muscle excitations that, in concert with the ground reaction forces, generated a forward dynamic simulation that tracked the fine-tuned kinematics generated by the residual reduction algorithm [19] [20]. Computed muscle activations were compared to the experimental EMG patterns so as to evaluate the fidelity of the simulation's muscle activation timing.

Tripping-perturbation simulation

Each previously created (unperturbed) forward-dynamic simulation was run using the computed muscle excitations and the experimental ground-reaction-force data while applying a time-invariant additional external force to the swing-foot of the model. This simulated the subjects' passive response to that tripping perturbation. The force was applied to the center of mass of the swing-foot's calcaneus segment, and was directed horizontally backward with a magnitude of 10% of the subject's body weight. This force was similar, but was not meant to be identical to observed backward forces between a tripping foot and an obstacle in an actual tripping of a human [21]. Time evolution of the tripping force was not considered here as the objective was to study stability against a force, similar in magnitude to a real tripping force, but not the actual tripping.

In a single tripping-perturbation simulation the equations of motion were integrated forward for a period of $t_t = 20$ ms. Note that the tripping-perturbation force was applied during the entire t_t . The initial state for each tripping-perturbation simulation was the state of the unperturbed simulation at the given instant of time. This analysis was repeated at 1 ms intervals throughout the swing phase. Since the data were limited to the swing phase [14], and the first 30 ms after the toe-off were needed to setup the simulation, the first tripping perturbation could be started after that time, and the last tripping perturbation at time t_t prior to the heel-strike. A batch of the tripping simulations was run by using a program written in Matlab (MathWorks, Inc., Natick, Massachusetts, USA), which automatically executed OpenSim [15]. The program for the batch simulations is freely available at www.simtk.org.

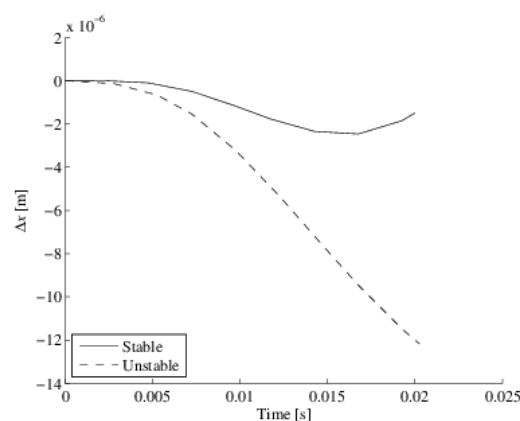


Figure 1. Examples of a stable and an unstable stage. A difference between the perturbed and unperturbed positions of the trunk as a function of time for a stable and an unstable stage.

For each tripping perturbation, a difference between a perturbed position, x_p , and an unperturbed position, x_u , of the trunk center of mass, in the fore-aft direction, as a function of time, t , was computed as follows:

$$\Delta x(t) = x_p(t) - x_u(t).$$

The positive direction of x is forward.

Each tripping perturbation was utilized to see if walking was stable at the respective instant of time. It has previously been shown that the magnitudes of responses to the tripping perturbation depend on the size of the subject [12].

Consequently, these magnitudes are not useful when comparing subjects of different size. A more universal measure is thus called for. Walking was defined as stable at a given instant of time if the following condition was met: an absolute difference between the perturbed and unperturbed positions of the trunk at the end of the perturbation was smaller than its absolute peak value during the perturbation. In other words, at a given instant of time, walking was defined as stable if the deviation of the position of the trunk due to the perturbation was bounded. Such a stable state means that after initially deviating from its unperturbed position, the trunk then approaches and stays close to that position. An unstable state means in turn that the trunk is moving away from its unperturbed position. Examples of a stable and an unstable stage are illustrated in Fig. 1.

The percentage of the swing phase during which the subject was stable was determined from all the performed perturbations. A group average of that percentage was calculated for each subject group. Student's t-test was performed to test whether a difference between the group averages is statistically significant.

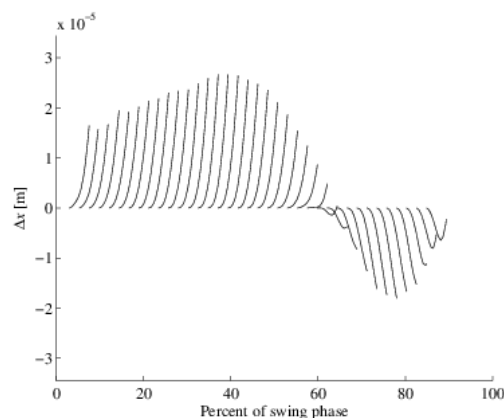


Figure 2. Responses to the tripping perturbations for one unimpaired subject. Each plotted curve represents a difference between the perturbed and unperturbed positions of the trunk as a function of the percentage of the swing phase. Every tenth perturbation performed during the swing phase is shown in the figure.

Results and discussion

Responses to the tripping perturbation are depicted for one unimpaired subject and for one CP subject in Figs 2 and 3, respectively. All unimpaired subjects were stable at some point of their swing phases, whereas six out of the ten CP subjects were never stable. The unimpaired subjects were statistically significantly stable for a larger percentage of the swing phase than the CP subjects ($p < 0.05$). On the average, unimpaired subjects were stable for 8.5% of the swing phase while CP subjects were stable for 1.5% of the swing phase. These findings disclose that the bulk of the swing phase is unstable even in unimpaired individuals. However, CP seems to be related to deteriorated proactive control of balance as the CP subjects were less able to stabilize their walking.

The results indicate that momentum induced by the perturbation is distributed over the body segments in a different fashion in CP subjects compared with unimpaired subjects. This difference likely stems from crouch gait and different muscle forces exhibited by individuals with CP [14]. Figures 2 and 3 demonstrate that responses to the tripping perturbation depend on timing of the perturbation. This may be related to an experimental finding that the timing of a tripping perturbation affects the selection of a recovery strategy [22].

The CP-subject group (age 8.1 ± 1.7 yrs) was slightly younger than the unimpaired-subject group (age 12.9 ± 3.3 yrs). This age difference is not, however, expected to explain the observed difference in stability because the youngest unimpaired subject (age 7 yrs) exhibited the largest percentage of stable swing phase.

It should be kept in mind that the results are trip-specific, other types of perturbation would likely produce very different characteristics. In light of the above, the results cannot be considered as a general or complete picture of walking stability. Regardless of this, trip-specific characteristics are important when studying trip-related falls.

A reactive control plays an important role in the prevention of falling. However, the effectiveness of a proactive control, considered here, determines the need for a reactive control. Furthermore, a proactive control enables better reactive control through stabilization of the optic flow and an effective processing of the vestibular-system signals as discussed above. Therefore, a proactive control contributes to the prevention of falling.

It is worth noting that simulations did not necessarily include all features of the subjects, which are related to their stability of walking. A direct validation of the muscle forces that result from simulation is not possible as the actual muscle forces of a human subject cannot usually be measured. This poses a challenge for creating completely subject-specific simulations. Nevertheless, a successful comparison of experimental and simulated data for normal

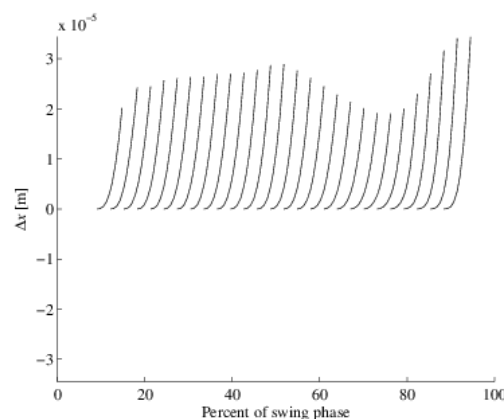


Figure 3. Responses to the tripping perturbations for one CP subject. Each plotted curve represents a difference between the perturbed and unperturbed positions of the trunk as a function of the percentage of the swing phase. Every tenth perturbation performed during the swing phase is shown in the figure.

gait of the subjects considered here has been done by showing that their simulated kinematics tracked the measured ones well and that simulated joint moments closely matched those determined in the experiments [13] [14]. The EMG data have shown some disagreement with the simulated muscle activations [13] [14]. Such disagreement may have been due, in part, to measurement errors in the experimental gait data. Also, the generic model does not account for subject-specific muscle-tendon properties.

Conclusions

Subjects with spastic diplegic cerebral palsy were stable for a smaller percentage of the swing phase than unimpaired subjects when subjected to a tripping perturbation. Furthermore, six out of the ten CP subjects were never stable, whereas all unimpaired subjects were stable at least during some interval of their swing phases. These findings demonstrate that a tripping perturbation is capable of detecting a balance impairment. Thus, it shows promise as a tool for rehabilitation.

Declarations




























Competing interests

The authors declare that they have no competing interests.

Authors' contributions

RK conceived the study, performed all analyses, interpreted the data, and drafted the manuscript. PM, JA, and JT interpreted the data and revised the manuscript. All authors read and approved the final manuscript.

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