J Braz. Soc. Mech. Sci. Eng. (2014) 36:491–500
DOI 10.1007/s40430-013-0055-5

TECHNICAL PAPER

Mechanical model of the recovery reaction from stumbling: effect of step length on trunk control

A. Forner-Cordero · HFJM Koopman ·
FCT van der Helm

Received: 14 December 2012 / Accepted: 23 June 2013 / Published online: 24 January 2014

Abstract Falling after a gait perturbation is a major problem for elderly people. The goal of this paper is to model some mechanical limitations of the recovery strategies performed after a trip or stumble, such as elevating or lowering strategies. A biomechanical model of the recovery was used to interpret stumbling data measured on healthy young and elder participants. The experiments consisted of simulating a stumble by blocking the swing leg, while the subject was walking on a treadmill. The motion and the vertical ground reaction forces were recorded to calculate the inverse dynamics. It is hypothesized that the stumble recovery depends on the ability to control the trunk. In the elevating strategy, the swing foot lands ahead of the body center of mass (COM) and the trunk flexion torque can be compensated. In the lowering strategy, with a shorter step, the trunk flexion cannot be arrested without the application of antero-posterior horizontal ground reaction forces. This action accelerates forwardly the COM, thus requiring quick successive recovery steps to place the foot ahead of the COM. If the recovery step is too slow, it is impossible to arrest trunk flexion and a fall would occur. The model and experimental data suggest that step length and speed after stumbling limit the recovery. Further research on screening protocols of maximal step speed is advised to evaluate the risk of falling in the elderly.

Keywords Gait · Falls · Stumble · Recovery strategy · Trip · Gait control

1 Introduction

The consequences of the fall are one of the most serious problems for elderly people. Most of these falls occur when the subject cannot recover from a perturbation during gait, such as a slip or a trip [1]. Stumbling, in which the swinging leg strikes an obstacle, was reported as a major contributor to falls causing up to 38 % hip fractures in elderly people. Several experiments have measured the stumble reaction during gait and described three recovery strategies according to the behavior of the perturbed (swinging) leg [2–5]. The choice of strategy depended on the perturbation instant and its duration: (a) elevating strategy that consists of an elevation of the swing limb to clear the obstacle and that occurs more frequently as a response to early swing perturbations, (b) lowering strategy that consists of bringing the foot to the ground as quickly as possible and that appears as a response to mid or late swing perturbations, (c) delayed lowering strategy in which the subject first tries an elevating strategy and then switches to

Technical Editor: Fernando Alves Rochinha.

A. Forner-Cordero (✉)
Biomechatronics Lab, Mechatronics Department, Escola Politécnica da Universidade de São Paulo, Av. Prof. Mello Moraes, 2231, Cidade Universitaria, Sao Paulo, SP 05508-030, Brazil
e-mail: aforner@usp.br
URL: http://sites.poli.usp.br/pmr/biomecatronica/; http://sites.poli.usp.br/p/arturo.forner/

HFJMKoopman
Institute for Biomedical Technology (BMTI), Biomechanical Engineering, Construerende Technische Wetenschappen, Universiteit Twente, P.B. 217, Gebouw Horstring Kamer W 212, 7500 AE Enschede, The Netherlands

FCTvan der Helm
Biomechanical Engineering Department, Faculty of Mechanical Engineering. Delft University of Technology, Mekelweg 2, 2628 CD Delft, The Netherlands

a lowering one, occurring typically in perturbations during early swing with long duration [6]. However, the description of the recovery strategies does not answer the question: why did the fall occur?

Several factors have been empirically associated with the success of the recovery: (a) to perform a quick appropriate reaction, (b) to control the forward rotation of the trunk [7], (c) to execute a step of sufficient length [8, 9], (d) the ability of the stance limb to support the body during the perturbation and after touchdown of the tripped leg, (e) to provide sufficient hip height during double stance to execute a follow-through step [10]. Three types of falls were described by Pavol et al. [10]. One type occurred during the execution of an elevating recovery strategy and it was associated with a faster walking speed followed by an excessive lumbar flexion. The other two types occurred during the lowering strategies and were named during-step and after-step falls. The during-step fall was associated with a faster walking speed and a delay in the stepping response. The after-step falls were related to a more anterior position of the center of mass (COM) at the instant of the trip followed by an excessive flexion of the trunk and buckling of the recovery limb. However, this experimental work lacked a biomechanical model to explain the relationship between the variables. The factors that affect the lowering strategy from a stumble in the elderly were analyzed with a model combined with experimental data [11]. It was concluded that the angle between the ankle joint and the body COM at the foot contact instant after the trip (body tilt angle) was a good indicator of the recovery success within the perturbed step. In addition, it was shown that the time of response had a larger impact on the body tilt angle than the gait speed. However, the tilt angle did not predict falls that occurred in the subsequent recovery steps, suggesting that other mechanisms influence the recovery. One of these mechanisms could be the control of angular momentum of the body after the trip [12].

This paper analyzes the mechanical limitations of the recovery strategies after a stumble using a biomechanical model that explains the role of the external forces on the control of the trunk and gait stability [13]. The model is used to analyze measured data from experimentally induced recovery reactions. It is hypothesized, as proposed by other authors, that the recovery from a gait perturbation can be described as an effort to control the forward flexion torque on the trunk [8, 10, 12]. This hypothesis is evaluated in terms of the mechanical limitations in the trunk control resulting from the application of ground reaction forces. It has been shown that a lowering strategy resulted in a shorter step and then a reduced ability to control the trunk during double stance [13]. The hip kinematics, ground reaction forces (GRF) and centers of pressure (CoP) depend on the step length, which limits the maximal trunk moments. Considering that the different recovery strategies result in distinct combinations of step lengths, hip positions and trunk angles, each strategy imposes certain limitations for the recovery.

2 Materials and methods

2.1 Model of the double stance after a stumble

A three-link model in the sagittal plane (see Fig. 1b) was used to model the subjects during the double-stance phase of gait [13]. It was shown experimentally that most of the energy exchange in the recovery reaction occurred during double stance [14]. The model presented here is inspired by previous models of gait, such as the simplest walker [15] and the simplest walker with trunk [13, 16]. During double stance, both legs are fully extended on the ground and it can be assumed that the acceleration of the legs is negligible and that there is no knee motion. Therefore, both legs can be regarded as rigid links between the respective centers of pressure on the ground and the hip joint. This model relates the hip moment M^z_{hip} (Eq. 1) that controls the trunk upright as an inverted pendulum to the positions of the hip relative to the CoP under each foot (trailing, right: x_{CoPR}; leading, left: x_{CoPL}) during double stance (Eq. 2):

$$\begin{aligned} M^z_{\mathrm{hip}} = {} & (I_{\mathrm{CT}} + a_{\mathrm{T}}^2 \times m_{\mathrm{T}}) \times \ddot{\theta}_{\mathrm{T}} + m_{\mathrm{T}} \times a_{\mathrm{T}} \\ & \times \left((\ddot{y}_{\mathrm{hip}} + g) \times \cos\theta_{\mathrm{T}} - \ddot{x}_{\mathrm{hip}} \times \sin\theta_{\mathrm{T}}\right) \end{aligned} \tag{1}$$

$$\begin{aligned} M^z_{\mathrm{hip}} = {} & F^x_{\mathrm{gR}} \times y_{\mathrm{hip}} + F^x_{\mathrm{gL}} \times y_{\mathrm{hip}} - F^y_{\mathrm{gR}} \times (x_{\mathrm{hip}} - x_{\mathrm{CoPR}}) \\ & + F^y_{\mathrm{gL}} \times (x_{\mathrm{CoPL}} - x_{\mathrm{hip}}) + m_{\mathrm{L}} \times g \\ & \times (b_{\mathrm{R}} \times \cos\theta_{\mathrm{R}} + b_{\mathrm{L}} \times \cos\theta_{\mathrm{L}}) \end{aligned} \tag{2}$$

The variables in this equation are, as described in Fig. 1b:

- CoPR and CoPL: center of pressure under trailing (right) and leading (left) legs;
- m_{T}, m_{L}, masses, respectively, of the head–arms–trunk and the leg;
- I_{CT} inertia moment of the head–arms–trunk with respect to the center of mass;
- a_{T}, b_{R} and b_{L} are the distances between the hip joint and, in that order, the center of mass (COM) of the load cell, ms and trunk, right leg and left leg, respectively;
- L_{R}, L_{L} lengths of the modeled legs between the hip joint and the CoP of each foot;
- x_{hip}, y_{hip} are the hip joint positions in antero-posterior (A-P) and vertical directions, respectively;
- x_{CoPR}, x_{CoPL} are the positions of the CoP under right (trailing) and left (leading) foot;
- F^x_{gR}, F^y_{gR} and F^x_{gL}, F^y_{gL} horizontal (A-P) and vertical GRF, correspondingly, on trailing and leading CoP;

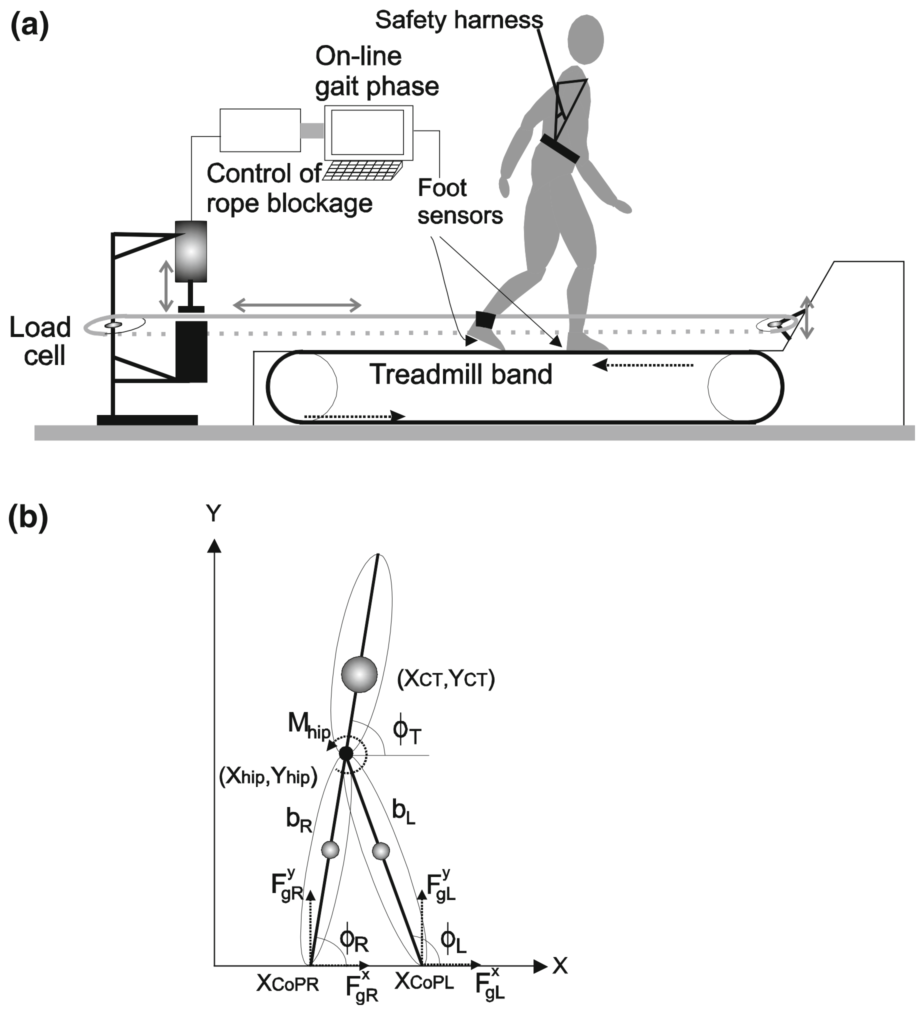

Fig. 1 **a** Drawing of the experimental setup to induce stumbling reactions on a treadmill. **b** Three-link model used for the recovery (see text): a_T, b_R and b_L represent the distances between the hip joint and, respectively, the COM of the trunk, right and left legs. x_{hip}, y_{hip} are the hip joint positions. F^x_{gR}, F^y_{gR} and F^x_{gL}, F^y_{gL} are the horizontal and vertical GRF, respectively, on right and left CoP and θ_T, θ_R and θ_L represent the segment angles of the trunk, right and left leg, respectively with respect to the horizontal

– θ_T, θ_R, θ_L segment angles of the trunk, right (trailing) and left (leading) leg, respectively with respect to the horizontal.

When the hip is between the centers of pressure (normal walking) the vertical forces on the leading and the trailing limbs contribute, respectively, to net extensor and flexor moments at the hip, allowing the control of trunk moments. If the hip is ahead of the leading CoP, e.g., during a perturbation, the vertical GRF cannot contribute to an extensor moment at the hip, and the trunk might fall forward. Any extensor moment at the trunk results in a forward acceleration of the hip (Eq. 2).

2.2 Experiments

Four healthy young and two healthy elderly male volunteers participated in the stumbling experiments wearing their own rubber-soled laced shoes (see Tables 1, 2). All of them walked normally on a treadmill after a maximum of 10 min of practice. No participant reported neurological or musculo-skeletal problems, joint replacements, cardiac or vestibular problems, recent injuries or history of falls. The experimental protocol was approved by the medical ethical committee of the local rehabilitation hospital.

2.2.1 Apparatus

The stumble perturbation was applied during treadmill walking and it was applied by blocking a rope attached to the left ankle, thus obstructing the forward swing of the leg (Fig. 1a). The rope was blocked with a brake. The brake was actuated by a pneumatic cylinder (AH-35-50, Festo) and it was equipped with a load cell (LM-20KA, Kyowa) to measure the perturbation force. The experimenter selected the perturbation instant (early, mid, late swing) in a Labview real-time program (National Instruments) that monitored the heel contacts with a force-sensitive resistor and triggered the pneumatic cylinder. The foot contacts were measured at 1 kHz by means of two footswitches on the foot sole at the heel and the head of the fifth metatarsal bone in both feet. The contacts were used to control the correct instant with respect to the gait cycle to block the

Table 1 Healthy young subject characteristics and recovery strategy chosen

Subject	Height (cm)	Weight (kg)	Age (years)	Elevating	Lowering	Delayed lowering	Fall
A	167	84	28	2	4	1	1
B	183	80	40	2	4	3	
C	181	67	22	2	4	7	
D	181	83	24		6	1	
Total				6	18	12	1

Table 2 Healthy elderly subject characteristics and recovery strategy chosen

Subject	Height (cm)	Weight (kg)	Age (years)	Gait speed (m/s)	Elevating	Lowering	Fall
E	172	86	70	1.1 (initial)			5
				0.86		2	1
				1	1	2	
				1.1 (final)	1	2	
F	179	82	77	0.56	2	3	
Total					4	9	6

rope. A chest harness attached by a dynamic rope to a frame prevented the subject from falling. The rope was loose while the participant walked; if the subject fell leaning on the harness, the rope became tensed and the subject could not continue walking. In the experiments with the elderly, a load cell (STS50KGC2, Sun Scale Inc.) placed in series with the safety rope measured if the subject was leaning on the rope. When the force reached 15 % or exceeded 10 % of the body weight within three recovery strides, the trial was discarded as a rope-assisted recovery.

2.2.2 *Experimental protocol*

The perturbation onsets were aligned at early swing with short (250 ms) or long (450 ms) durations and at mid and late swing only with short durations. The perturbation order and the time between them were random. The subject was continuously walking on the treadmill without knowing if a perturbation was going to occur.

The reference walking speed was 1.1 m/s and it was reduced when the subject could not walk comfortably. If a participant fell (leaning on the harness) up to five consecutive times due to the perturbations, the speed was reduced in subsequent trials until the subject was able to recover. If the participant recovered more than five times at a lower gait velocity, the treadmill speed was increased again up to the reference or to the maximal speed at which the subject could recover.

2.2.3 *Measurements*

The motion of the body was measured at 50 Hz by means of a five-camera optical system (VICON 370. Oxford Metrics, UK). The VICON optical measurement system used five cameras with infrared light-emitting diode strobes to record the three-dimensional trajectories of retro-reflective markers placed at specific anatomical locations (modified Helen Hayes' protocol described in [17]). The same steps were recorded at 50 Hz with instrumented insoles Pedar® (Novel, gmbh). These insoles consist of an array of pressure sensors that can be placed inside the user shoes. With the motion data, the vertical GRF and the CoP, the complete three-dimensional GRF were estimated following optimization procedures described elsewhere [17–19].

2.3 Data analysis

The foot contact times were obtained from the vertical GRF measured with the insoles. The distance between both centers of pressure (CoP) at heel strike defined the step length and the step time was the interval between consecutive heel strikes of different limbs. The step speed was defined as the ratio of step length and time. The recovery strategies were clustered in three groups according to the normalized step length and time (K-means clusters, SPSS Inc.). The elevating strategies resulted in normal gait steps lengths with longer durations, while the delayed lowering ones had shorter steps.

The lowering strategies had shorter step length and time (Fig. 2b), in agreement with previous research [3–5].

The hip margins are the distances between the hip joint and both CoP (X_{CoPR} and X_{CoPL}) in the antero-posterior (A-P) direction at heel strike and describe, according to the model, the boundaries of the possible hip torques. The body tilt was defined by the angle between the vertical and the line joining the estimated position of the CoM (trunk)

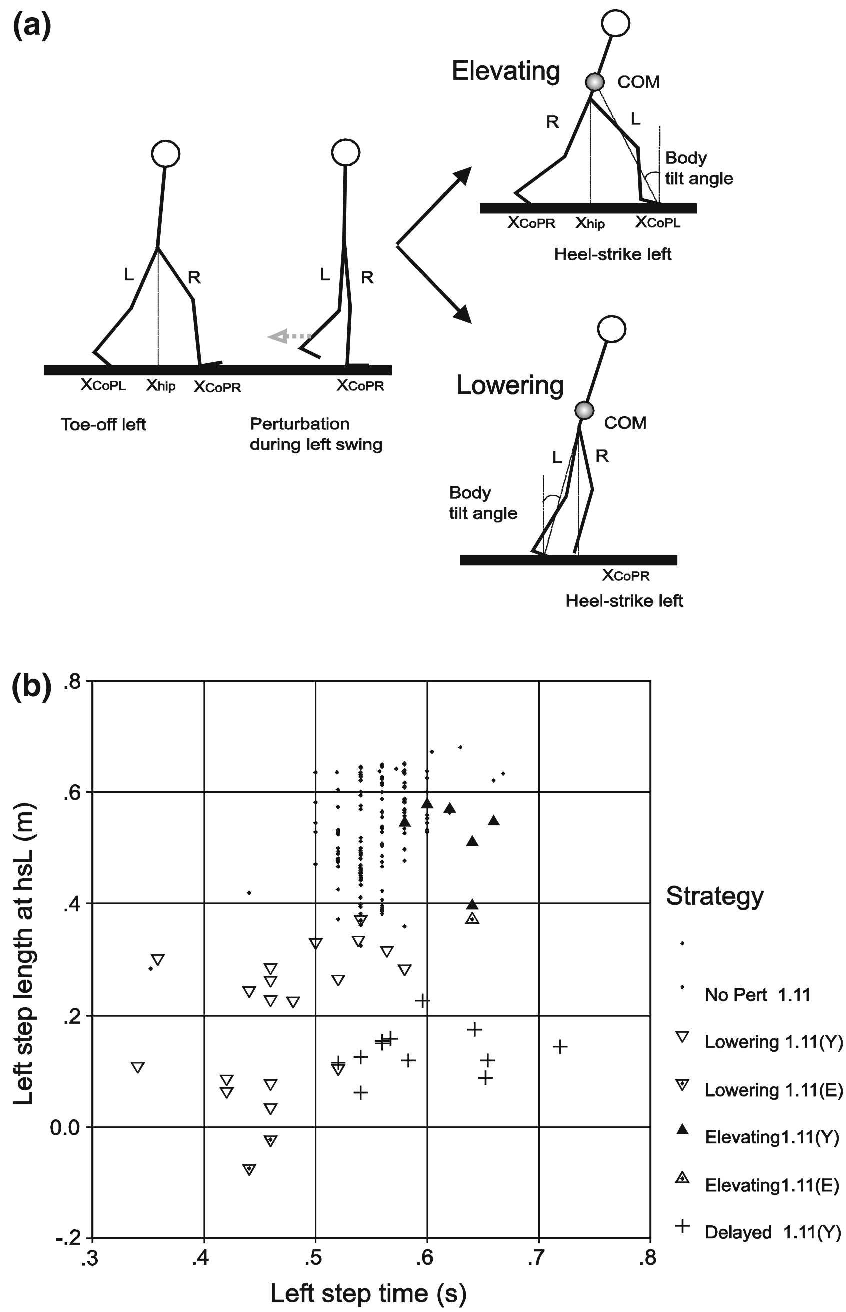

Fig. 2 **a** Stick diagram representation of the perturbation and possible recovery reactions. With the elevating strategy, the swing leg (X_{CoPL}) lands in front of the COM, resulting in a negative body tilt angle, a normal step length and prolonged step time. With the lowering and delayed lowering strategy, the swing leg lands behind the COM, resulting in a positive body tilt angle, a short step length and a shortened (lowering) or prolonged (delayed lowering) step time. **b** Perturbed left step length (distance between CoP at heel strike, left) and step time (time from the heel strike, right, to the next heel strike, left) with the corresponding classification according to the strategy at 1.1 m/s. The normal gait reference values (no Pert) are included in the graph

and the center of pressure of the leading (left) limb (X_{CoPL}) at hsL after the perturbation (Fig. 2a). The complete GRF, joint angles, hip positions and the CoPs at heel strike and toe off were retained for further analysis.

3 Results

The young volunteers walked on the treadmill at 1.1 m/s. One subject fell at the first perturbation.

For the eldest participant, the maximal gait speed was 0.56 m/s and no fall occurred. The other elder participant walked normally at 1.1 m/s. However, he could not recover from perturbations, and after five falls the speed was reduced to 0.86 m/s. After five recoveries in each condition, the speed was increased to 1 m/s and finally to 1.1 m/s because the subject did not fall again. The classification of strategies as a function of the step length and time resulted in Fig. 2b.

3.1 Analysis of the perturbed stride and the body tilt

The hip position relative to X_{CoPR} (stance limb) was more advanced after the perturbation for delayed and lowering strategies than for the elevating strategy in the experiments with healthy young (Fig. 3).

The hip margins in the elevating strategies and normal gait were similar. However, the hip margins decreased for the lowering and delayed lowering strategies, leading to different combinations during double stance. In most of the cases, the landing occurs with the hip between both CoP (Fig. 3). It was expected that if the hip was ahead of both X_{CoPL} and X_{CoPR} a fall would result. The recoveries in this situation occurred at a lower walking speed. However, when the hip was behind the trailing leg X_{CoPR} (stance limb) at hsL, there was always a fall.

The dashed line in Fig. 3 shows the line in which both centers of pressure are at the same antero-posterior position. This line discriminates between the falls and the non-falling cases for the higher speed (1.1 m/s), because the few cases in which the subject recovered above this line were measured at lower speed.

The body tilt angle at heel strike after the perturbed swing (Fig. 4) showed remarkable differences between strategies suggesting that it had an important role in the recovery.

In most of the falls, the body tilt value exceeded 10° (Fig. 4). The two recoveries with tilt angles larger than 10° corresponded to a lower walking speed (1 m/s). The body tilt angle and the step velocity were highly correlated ($R = 0.965$).

3.2 Horizontal ground reaction forces

At the end of the double stance after the perturbation (toR: right toe off), a linear relation between the positions of the hip with respect to the CoPL and the A-P ground reaction forces (GRF) was discovered ($R = 0.758$ in Fig. 5). For normal walking and elevating strategies, the A-P GRF were negative, but their values increased as the hip margins decreased, reaching positive values for small hip margins (Fig. 5).

The horizontal ground reaction forces reached positive (forward) values when the hip was ahead of CoPL. With the hip ahead of both CoPL and CoPR, the horizontal GRF under each foot should be positive, resulting from an extensor hip moment.

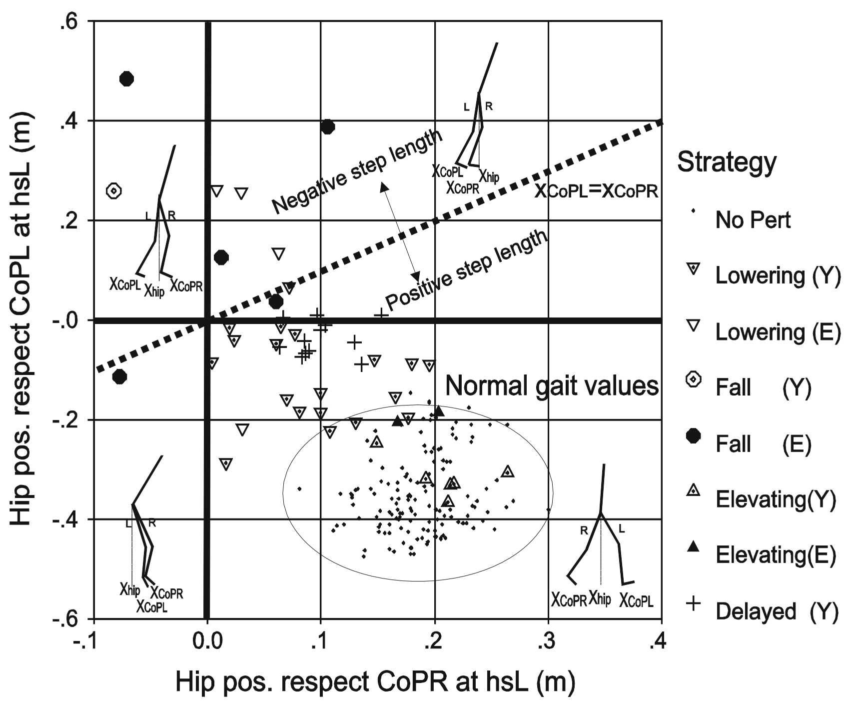

Fig. 3 Hip positions with respect to the leading (*left*: CoPL) and trailing (*right*: CoPR) limbs at the heel strike, left, after the perturbation. As a reference, the normal gait values (no Pert) are included: COM in front of CoPR (*positive values*) and trailing CoPL (*negative values*). In the elevating strategy and most of the delayed lowering cases, the normal situation is restored. In the lowering strategy, the COM is ahead of CoPR and CoPL, necessitating a quick second step, but often resulting in a forward fall. If the COM remains behind the CoPR, a backward fall is unavoidable. The results are classified by strategy with a distinction between elderly (E) and young (Y). Only the trials measured at speed higher than 0.8 m/s have been included. The *dashed line* shows the line in which both CoPR and CoPL are at the same antero-posterior position. The *stick diagrams* represent the two most representative body configurations in the corresponding hip margin space

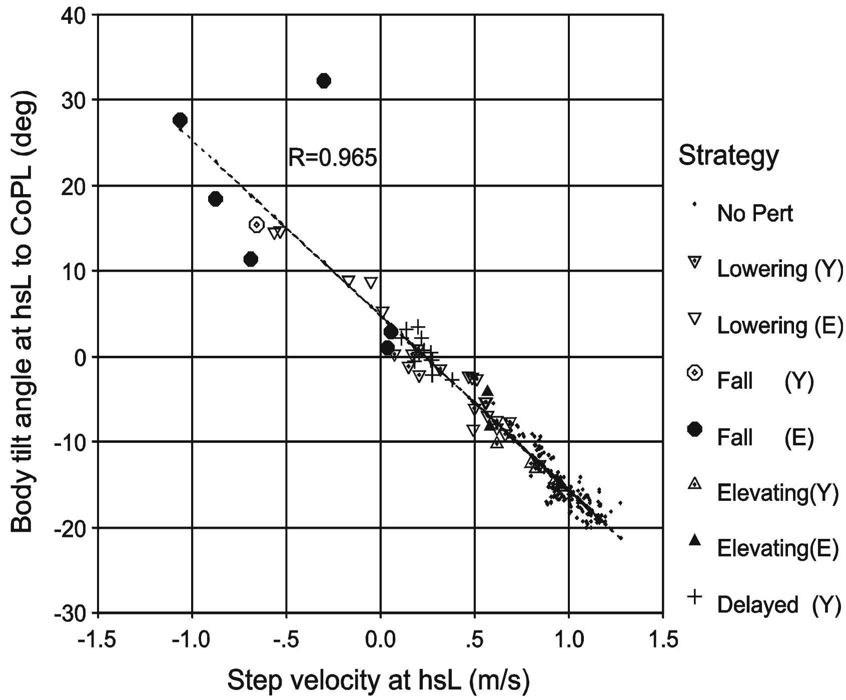

Fig. 4 Body tilt angle versus velocity of the perturbed step: lower step velocities result in positive body tilt angles. The results are classified by strategy with a distinction between elderly (E) and young (Y). Only the trials measured with gait speed higher than 0.8 m/s were included. The regression line (predicted values of body tilt) and the correlation coefficient *R* are also included. The linear model obtained was: body tilt $= -20.5 \times$ step velocity $+ 4.8$

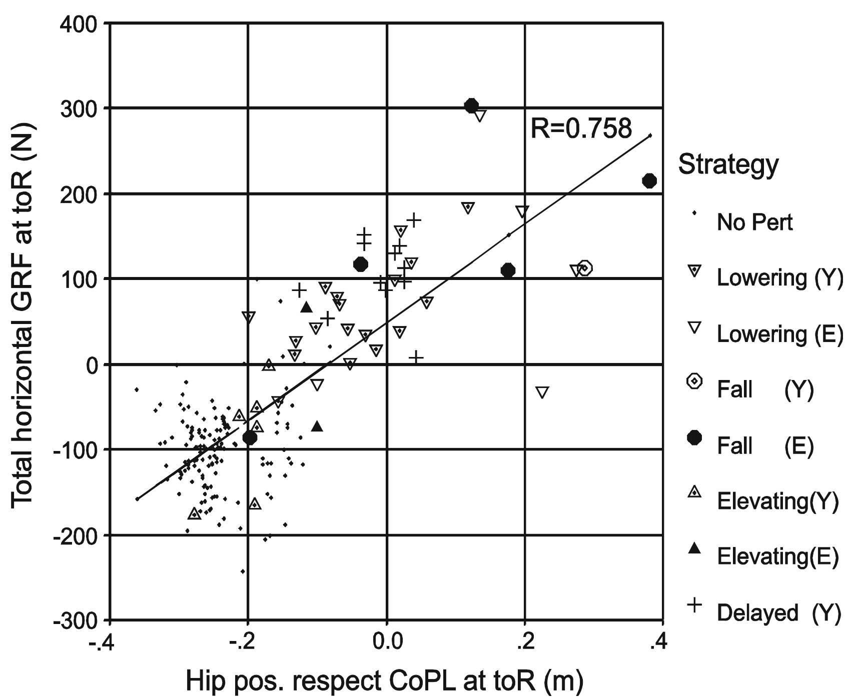

Fig. 5 Scatter plot of the hip position with respect to the leading limb (CoPL) and the total horizontal GRF at the end of the double stance after the perturbation (toR). The responses are classified according to the strategy with a distinction between elderly (E) and young (Y). Only the trials with a gait speed higher than 0.8 m/s were included. Normal walking (no Pert) and the falling cases (fall) are also included. The regression line (predicted values of total horizontal GRF at toR) and the correlation coefficient *R* are also included. The linear model obtained had a slope of 575.201 with a constant of 48.387. Two outliers with horizontal forces larger than 400 N were removed from the graph

4 Discussion

This paper underscores the major role of the GRF in controlling the trunk during gait. This role is shown with the model and confirmed with the gait perturbation experimental results. One of the consequences of a trip or stumble during gait was the reduction of the recovery step length, revealing the impossibility of controlling the trunk flexion–extension without horizontal GRF. These forces accelerate forward the body COM that would require a quicker step to recover balance. It has been reported that the control of the trunk was crucial in the recovery from a stumble; however, it was also found that the trunk extensor muscles did not affect recovery performance [7]. These contradictory results could be explained because feet placement restricts the trunk control more than trunk extensor musculature. The step legth limits the maximal trunk torques, independently of other possible limitations such as maximal muscle forces, lower limb strength and arm coordination that were proven significant factors in the recovery [12, 13, 20–22].

The individual strategies to recover from a stumble have been analyzed with a model and experimental stumbling data. In the elevation strategy, a normal gait step length was obtained with a prolonged step time (Fig. 2b), resulting in the swing foot landing in front of the hip joint centers (Fig. 3). In such case, a hip extension moment can be generated and a full recovery is achieved within one step. In the lowering and delayed lowering strategies, the foot lands behind the hip joint center and the COM. To obtain an extensor moment in the trunk, a positive horizontal GRF is required (Eq. 2; Fig. 5), accelerating forward the body COM and requiring fast subsequent steps to catch up. If these are not fast enough, a fall will occur.

According to the model (Eq. 2), the limits for maximal hip torques are very narrow for short step lengths (lowering strategy) and the horizontal forces come into play to generate trunk extension moment (Fig. 5). However, too large horizontal forces would accelerate the body COM requiring a subsequent step quick enough to recover.

The model presented here has some simplifications that have to be considered. From a mechanical point of view, it is a bi-dimensional model, so it is not possible to compute any other torques than the flexo-extension (sagittal plane). The data measured experimentally were three dimensional and the complete forces and moments were computed at each joint, as described in [19]. However, the step behavior explained by the model agrees with the found experimental data; therefore, it was chosen to keep the model as simple as possible. Nevertheless, it would be very interesting to include a three-dimensional mechanical model in a future work. Another limitation of the model is the assumption that the legs are fully extended and behave like rigid bodies, thus, neglecting knee flexion. Although the knee joint behaves like a high stiffness spring during stance [23], it has been shown that a model with compliant leg behavior can reproduce better the stance dynamics [24]. Such a model would introduce additional complexity that is not needed to support the conclusions of this paper. Nevertheless, including a compliant leg would be a natural extension of this work to obtain further insight into the gait perturbation recovery limitations.

4.1 Interpretation of the experimental results

The kinematics of the observed recovery responses of these six subjects agree with previous experiments that measured similar tripping and stumbling perturbations during gait [2–5, 12]. In this respect, although the experimental sample was not large it can be regarded as adequate to assess the model [13].

The horizontal GRF increased with the hip margin reduction with respect to X_{CoPL} as predicted in Eq. 2 (Fig. 5). This original finding suggests an important aspect in the control of gait. The horizontal GRF are minimized, probably to avoid unnecessary accelerations of the body COM that might cause an energetically inefficient gait. Nevertheless, stability requirements overrule this energetic requirement, as shown in Fig. 5; if the step was too short, larger horizontal GRF appear to arrest the trunk forward flexion, in agreement with the model. In addition, this mechanism has also been described experimentally when subjects had to shorten their step during gait [25].

Several recovery steps from early swing perturbations in the elderly were unsuccessful and resulted in a fall. These steps had a negative length and the hip was ahead of the X_{CoPL}, indicating that the foot was placed on the ground before overtaking the stance leg at mid-swing. This situation did not occur with the younger subjects because they performed elevating strategies for early swing perturbations [3]. In this strategy, the support limb must provide time and clearance for adequate recovery limb positioning with sufficient hip margins during double stance [11, 13].

Therefore, most of the “falls” in the experiments occurred when the hip was ahead of the X_{CoPL} (Fig. 3). It seems that it was impossible to counteract the forward trunk flexion and the subject could not make a step quick enough to recover. This mechanism may explain the during-step and the after-step fall mechanisms described previously [10]. In the first case, the during-step fall, the GRF position makes it impossible to counteract the forward-falling flexion of the trunk during the recovery step. In the second case, the after-step fall, the forward flexion of the trunk has been delayed by accelerating forwardly the hip joint with the application of antero-posterior GRF. However, these GRF accelerate forward the COM and the subsequent step must be quicker to allow a proper placement of the GRF application point (CoP). This mechanism would explain why the fallers who were walking faster and had slower steps fell significantly more frequently “during step”. Slower walking patients would have some margin to accelerate the hip forward generating a trunk extension moment to arrest the trunk flexion after the trip, but if the subsequent step was not quick enough they would have an “after-step fall” [10]. The body tilt angle discriminated between falls and recoveries in a group of elderly people performing lowering strategies [11]. The body tilt angle seems to be a good predictor of the recovery success and is highly correlated with the step velocity (Fig. 4), underscoring the importance of the step speed to avoid falling.

4.2 Configuration of the body at the end of the swing phase

During the double-stance phase of normal gait, the hip is always between both CoP (Fig. 3). This configuration allows controlling the trunk moments without accelerating the COM in the A-P direction [13]. The largest GRF component is the vertical one and it contributes to the hip moment multiplied by the hip margins (Fig. 3). In addition, the weight transfer from the trailing to the leading limb and the horizontal forces will characterize the possible moments to control the trunk. Therefore, it is possible to define different configurations during the recovery double stance:

- Inverted double stance: the weight-accepting limb (leading, left) is behind the limb that is being unloaded (trailing) and the hip, $x_{\mathrm{CoPL}} \leq x_{\mathrm{hip}} < x_{\mathrm{CoPR}}$. In Fig. 3, it is the region above the dashed line.
- False double stance: the hip is ahead of the stance limb $x_{\mathrm{hip}} > x_{\mathrm{CoPL}}, x_{\mathrm{CoPR}}$. Although both legs are on the ground, the ability to control the trunk is as constrained as during the single stance phase. In Fig. 3, it is the region of negative hip positions with respect to X_{CoPR}.

None of these configurations allows a complete control of the trunk moments, except if it would be possible to apply pull forces from the ground, as if the feet were glued to the floor. Figure 3 shows that all the cases of inverted double stance resulted in a fall.

The recovery from a false double stance can be achieved by the application of horizontal ground reaction forces. When both CoP are behind the hip, there is a net positive A-P GRF that contributes to the trunk extension (Eq. 2) accelerating the hip joint and the body COM (Eq. 1). The key point was to ensure that the next step would be quick enough to overtake the hip and provide enough hip margins to compensate the forward flexion of the trunk. The results presented here suggest that the ability to perform a quick step after stumbling to place the leading foot CoP ahead of the hip represents a limitation to a successful recovery.

4.3 Is the maximal step speed a limitation to recover?

It appears that one of the goals is to control the trunk forward movement and the model indicates that a necessary condition to accomplish this goal is an adequate feet placement. A similar mechanism based on the control of step length was described to control the stability of a biped walking robot [9].

The results presented here suggest that the maximal walking speed of a person at risk of falling should be limited by the maximal step speed to recover from the perturbation. Therefore, a measure of the ability to recover from a stumble could be based on the ability to perform quick steps with respect to the walking speed. In this respect, a large-scale clinical study may determine the falling risks in different elderly populations. In this context, it must be taken into account that the ability to perform the recovery reactions depends on other factors, such as neuromuscular or psychological factors, that are out of the scope of this work but pave the way to further multidisciplinary research. Finally, the biomechanical model presented here may be useful in pointing at the relevant variables and providing a basis to interpret the data.

5 Conclusions

This model and experimental results underscore the important role of feet placement and the horizontal antero-posterior GRF to control gait stability and recover from perturbations. During normal walking we use sufficient hip margins, the intervention of the horizontal GRF is minor and they hardly affect the COM accelerations. When a perturbation occurs and a lowering or delayed lowering strategy emerges, the hip margins are reduced (Fig. 3), the body tilt increases and the resultant GRF comes into play to stabilize the trunk. As shown in Fig. 5, there is a linear relation between the hip margins and the horizontal GRF. These results confirm the functional behavior described by the model (Eq. 2) to control the trunk movement.

References

1. Winter DA (1995) Human balance and posture control during standing and walking. Gait Posture 3:193–214
2. Dietz V, Quintern J, Berger W (1986) Stumbling reactions in man: release of a ballistic movement pattern. Brain Res 362:355–357
3. Eng JJ, Winter DA, Patla AE (1994) Strategies for recovery from a trip in early and late swing during human walking. Exp Brain Res (Experimentelle Hirnforschung, Experimentation Cerebrale) 102:339–349
4. Schillings AM, Van Wezel BM, Duysens J (1996) Mechanically induced stumbling during human treadmill walking. J Neurosci Methods 67:11–17
5. Forner Cordero A, Koopman B, van der Helm FCT (2003) Multiple-step strategies to recover from stumbling perturbations. Gait Posture 18:47–59
6. Schillings AM, van Wezel BM, Mulder T, Duysens J (2000) Muscular responses and movement strategies during stumbling over obstacles. J Neurophysiol 83:2093–2102
7. Grabiner MD, Feuerbach JW, Jahnigen DW (1996) Measures of paraspinal muscle performance do not predict initial trunk kinematics after tripping. J Biomech 29:735–744
8. Grabiner MD, Koh TJ, Lundin TM, Jahnigen DW (1993) Kinematics of recovery from a stumble. J Gerontol 48:M97–M102
9. Wisse M, Schwab AL, van der Linde RQ, van der Helm FCT (2005) How to keep from falling forward? IEEE Trans Rob V21(3):393–401

10. Pavol MJ, Owings TM, Foley KT, Grabiner MD (2001) Mechanisms leading to a fall from an induced trip in healthy older adults. J Gerontol Ser A Biol Sci Med Sci 56:M428–M437
11. van den Bogert AJ, Pavol MJ, Grabiner MD (2002) Response time is more important than walking speed for the ability of older adults to avoid a fall after a trip. J Biomech 35:199–205
12. Pijnappels M, Bobbert MF, van Dieen JH (2004) Contribution of the support limb in control of angular momentum after tripping. J Biomech 37:1811–1818
13. Forner Cordero A, Koopman B, Helm FCT (2004) Mechanical model of the recovery from stumbling. Biol Cybern 91:212–221
14. Forner Cordero A, Koopman HFJM, van der Helm FCT (2005) Energy analysis of human stumbling: the limitations of recovery. Gait Posture 21(3):243–254
15. Garcia M, Ruina A, Chatterjee A, Coleman M (1998) The simplest walking model: stability, complexity, and scaling. ASME J Biomech Eng 120(2):281–288
16. Wisse M, Schwab AL, van der Helm FCT (2004) Passive dynamic walking model with upper body. Robotica 22(6): 681–688
17. Koopman B, Grootenboer HJ, de Jongh HJ (1995) An inverse dynamics model for the analysis, reconstruction and prediction of bipedal walking. J Biomech 28:1369–1376
18. Forner Cordero A, Koopman HFJM, van der Helm FCT (2004) Use of pressure insoles to calculate the complete ground reaction forces. J Biomech 37:1427–1432
19. Forner-Cordero A, Koopman B, van der Helm FCT (2006) Inverse dynamics calculations during gait with restricted ground reaction force information from pressure insoles. Gait Posture 23(2):189–199
20. van Dieën JH, Pijnappels M, Bobbert MF (2005) Age-related intrinsic limitations in preventing a trip and regaining balance after a trip. Saf Sci 43(7):437
21. Pijnappels M, Reeves ND, Maganaris CN, van Dieen JH (2008) Tripping without falling; lower limb strength, a limitation for balance recovery and a target for training in the elderly. J Electromyogr Kinesiol 18:188–196
22. Krasovsky T, Baniña MC, Hacmon R, Feldman AG, Lamontagne A, Levin MF (2012) Stability of gait and interlimb coordination in older adults. J Neurophysiol 107(9):2560–2569
23. Cullell A, Moreno JC, Rocon E, Forner-Cordero A, Pons JL (2009) Biologically based design of an actuator system for a knee–ankle–foot orthosis. Mech Mach Theory 44(4):860–872
24. Geyer H, Seyfarth A, Blickhan R (2006) Compliant leg behaviour explains basic dynamics of walking and running. Proc R Soc B 273(1603):2861–2867
25. Patla AE, Robinson C, Samways M, Armstrong CJ (1989) Visual control of step length during overground locomotion: task-specific modulation of the locomotor synergy. J Exp Psychol Hum Percept Perform 15(3):603–617